EXERCISE STRESS CARDIAC MAGNETIC RESONANCE

DISSERTATION

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By

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ABSTRACT

This work presents the development and validation of a new cardiac stress imaging modality combining treadmill exercise stress testing with cardiac magnetic resonance imaging (exercise CMR). CMR offers several distinct advantages over standard stress imaging modalities. It has higher spatial resolution compared to nuclear scintigraphy without exposure to ionizing radiation, provides better contrast-to-noise than echocardiography, and offers comprehensive diagnostic information by combining stress cardiac function, stress myocardial perfusion, and myocardial viability. Although pharmacologic stress CMR has recently seen increasing clinical utilization, it has important drawbacks compared to exercise stress because the additional information derived from exercise adds to the diagnostic and prognostic value of a stress imaging test. This includes exercise capacity, blood pressure response, development of arrhythmias, and the presence or absence of symptoms such as chest pain during exercise. However, technical challenges with MR-compatible exercise and patient monitoring equipment as well as imaging in the presence of deep breathing and high heart rates at peak exercise stress have previously prevented the realization of exercise CMR.

The first step was to test the concept feasibility, combining cardiac wall motion and myocardial perfusion obtained immediately post-exercise with myocardial viability
at rest. The feasibility testing was successfully performed in 20 healthy subjects using a partially MRI-compatible treadmill located in the corner of the MRI room. However, several technical challenges including image artifacts and temporal resolution, accuracy of 12-lead electrocardiogram (ECG) monitoring, and the need for a fully MRI-compatible treadmill were identified.

Imaging optimization was carried out to compare the TSENSE and TGRAPPA methods of parallel imaging under the condition of deep breathing encountered after exercise stress, and both quantitative and qualitative results showed that TGRAPPA delivered superior artifact performance. TSENSE suffered from artifacts when the FOV was smaller than the object or the coil map was imperfect due to breathing motion. In addition, it was shown that using a 32-channel cardiac array coil enabled rate 4 parallel acceleration, resulting in improved temporal resolution.

Another aspect of exercise CMR to be considered is 12-lead ECG monitoring inside the MRI room, which is required both during treadmill exercise and recovery. Although the ECG is known to be non-diagnostic within the bore of any high-field magnet due to the magnetohydrodynamic (MHD) effect, the magnetic field threshold below which accurate ECG monitoring is feasible inside the MRI room but outside of the magnet bore was determined. It was shown that reliable ECG measurements could be obtained within the ST segment at magnetic field strengths below approximately 70 mT measured at the aortic arch. This corresponded to approximately 80 cm from the bore entrance for the Siemens 1.5T Avanto system, but could be extrapolated to any other

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system knowing the magnetic field plot. Based on this threshold, it was shown that accurate 12-lead ECG monitoring is feasible during treadmill exercise immediately adjacent to the magnet and during supine recovery from exercise on the MRI patient table.

Next, the feasibility of exercise CMR for accurate diagnosis of ischemia was investigated in 43 patients with known or suspected coronary artery disease who were referred for treadmill nuclear imaging. Both exercise CMR and nuclear data were obtained while exercising the patient only once. It was shown that exercise CMR could accurately detect coronary artery disease compared to coronary angiography as the gold standard, and preliminary results indicated favorable accuracy compared to nuclear stress imaging. However, the sample size would need to be increased in order to draw statistical conclusions. Successfully demonstrating that exercise CMR is diagnostically and prognostically superior to nuclear stress imaging in a larger clinical trial could have a significant impact on clinical practice and potentially change the current standard of cardiac stress testing.

Although these preliminary studies were conducted using the partially MRI-compatible treadmill in the corner of the MRI room, this would be impractical as a standard clinical stress imaging modality. It was necessary to minimize the time between end of exercise and imaging in order to capture rapidly resolving exercise-induced cardiac wall motion abnormalities, necessitating treadmill placement beside the MRI table. Requiring the patient to walk from a treadmill positioned any distance from the
MRI table would create a potential safety concern since patients may become lightheaded and subject to falling immediately following maximal exercise. Furthermore, there was risk of operator error in moving the treadmill too close to the magnet, and the entire setup in the corner of the room would not be possible at higher field strengths such as 3T. Therefore, a fully MRI-compatible water hydraulic treadmill was developed, which could be positioned immediately adjacent to the MRI table. This thesis presents all aspects pertaining to the development and testing of the feedback control system for continuous control of treadmill speed and elevation inside the MRI room. Component selection, design of speed and elevation control, LabVIEW software design and implementation, failure modes and effects analysis, and implementation of safety features are described. The results of speed and elevation performance testing are presented, followed by testing in healthy subjects and cardiac patients undergoing the standard Bruce treadmill protocol to reach peak stress immediately adjacent to the MRI table.

Finally, potential future directions beyond the detection of coronary artery disease are presented. These include right heart dysfunction, timing of valve replacement for Tetralogy of Fallot patients, physiologic studies of the relationship between wall motion and perfusion, and investigation of the mechanisms underlying myocardial fibrosis in elite endurance athletes.
DEDICATION

To Mark and to my family
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CHAPTER 1

INTRODUCTION

The aim of this thesis is to present the development and validation of a new imaging modality for detecting cardiovascular disease, which combines treadmill exercise stress testing with cardiac magnetic resonance (CMR) imaging. This new modality is referred to as exercise CMR. Chapter 2 presents the background behind exercise CMR, starting with an introduction to coronary artery disease (CAD), which is one of the principal applications of this technique. The importance of stress imaging and the limitations of the standard stress imaging modalities echocardiography and nuclear scintigraphy are discussed, followed by the potential advantages of CMR. The exercise CMR requirements, including the necessity to position the exercise device on or immediately adjacent to the MRI table in order to commence imaging as quickly as possible post-exercise and capture the exercise-induced cardiac wall motion abnormalities, are described. The case for why treadmill exercise is preferred to bicycle ergometry is presented. This is followed by a discussion of the challenges in operating a treadmill within strong magnetic fields, and the challenges in imaging in the presence of rapid heart rates and deep breathing encountered immediately post-exercise.
Chapter 3 describes a concept feasibility study for exercise CMR, combining cardiac wall motion and myocardial perfusion obtained immediately post-exercise with myocardial viability obtained after recovery from exercise. The results of feasibility testing in healthy subjects using a partially MRI-compatible treadmill located in the corner of the MRI room are presented, and the remaining technical challenges for CMR to become fully clinically realizable are identified. In Chapter 4, imaging optimization is performed to minimize artifacts and improve the temporal resolution. The TSENSE and TGRAPPA methods of parallel imaging are compared under the condition of deep breathing, and the use of a 32-channel cardiac array coil for enabling higher temporal resolution is investigated.

Another challenge with exercise CMR is electrocardiogram (ECG) monitoring, which is required both during treadmill exercise and recovery. Although the ECG is non-diagnostic within the bore of any high-field magnet due to magnetohydrodynamic effects, the magnetic field threshold below which accurate ECG monitoring is feasible inside of the MRI room is determined in Chapter 5. This threshold can be converted to the distance from the bore entrance knowing the magnetic field plot of any MRI system.

In Chapter 6, the feasibility of exercise CMR for accurately diagnosing ischemia is investigated in 43 patients with known or suspected CAD who were referred for treadmill nuclear scintigraphy. Based on these preliminary results, an estimation of the accuracy of exercise CMR is determined.
Chapter 7 presents the development and testing of the feedback control system for continuous control of speed and elevation of a fully MRI-compatible hydraulic treadmill which can operate immediately adjacent to the MRI table. The component selection, the design of speed and elevation control, the development of control software in LabVIEW, and the failure modes and effects analysis (FMEA) and the implementation of safety features are described. Next, the results of treadmill performance testing are presented, followed by testing in healthy subjects and cardiac patients undergoing the standard Bruce treadmill protocol to reach peak stress immediately adjacent to the MRI table.

Chapter 8 presents the conclusions and future directions, including new avenues for clinical research which were not previously available due to the limitations of existing stress imaging modalities. Examples include right ventricle dysfunction, the timing of pulmonary valve replacement in patients with Tetralogy of Fallot, physiologic studies of the relationship between wall motion and perfusion, and investigation of the mechanisms of myocardial fibrosis in endurance athletes.
CHAPTER 2

BACKGROUND

2.1 Introduction to Coronary Artery Disease and its Detection

Over 80 million Americans have one or more types of cardiovascular disease (CVD), the underlying cause of 34% of all deaths [1]. Approximately 54% of those deaths are specifically due to coronary artery disease (CAD). CAD affects approximately 16.8 million Americans with an estimated economic burden of $165.4 billion [1]. It is caused by the fatty build-up of plaque that leads to stenosis (narrowing) of the coronary arteries supplying the heart with oxygen (Figure 1). The plaque consisting of cholesterol, macrophages, foam cells, calcium, and other cell debris, is typically encapsulated within a fibrous cap. Rupture of the fibrous cap may activate a clotting cascade in an attempt to repair the injury, and result in the formation of a blood clot at the site of the rupture. This clot can suddenly block all blood supply to a part of the myocardium, resulting in a myocardial infarction.
Under normal resting conditions, the body can maintain a normal supply of blood to the heart through a stenotic coronary artery, such that no ischemia is evident at rest. As a result, stress testing of patients is used to detect coronary artery narrowing not apparent in a normal resting state. Under stress, the coronary blood flow in partially blocked arteries cannot increase sufficiently to meet the elevated metabolic demand of the myocardium. Regional differences in blood supply can be detected by perfusion imaging methods. Additionally, cardiac myocytes with an insufficient supply of oxygen have impaired function, resulting in detectable regional cardiac wall motion abnormalities. The physiologic response to exercise stress is further discussed in Section 1.2.

Figure 2 illustrates the heart and the coronary arteries (left anterior descending (LAD), circumflex (LCX), and the right coronary artery (RCA)) as well as the buildup of atherosclerotic plaque which impedes coronary blood supply to the myocardium. In order to standardize wall motion and perfusion analysis, the American Heart Association (AHA) has issued a set of guidelines for segmenting the heart [3], known as the 17-segment model. Each of the 17 myocardial segments are assigned to belong to one of
three coronary artery perfusion territories as shown in Figure 3. The segments are typically rated as normal, hypokinetic, akinetic, or dyskinetic based on wall motion [4], and normal or hypoperfused based on perfusion [5].

Figure 2: Illustration of the heart, the coronary arteries, and atherosclerotic plaque [2]

Figure 3: Seventeen-segment model of the left ventricle [3]
2.2 Physiologic Response to Exercise

Strenuous exercise is associated with a number of physiologic changes, including an increase in cardiac output (CO) from a resting value of 5 L/min to as much as 35 L/min [2]. CO can be expressed as a product of heart rate (HR) and stroke volume (SV), both of which increase during exercise:

\[ \text{CO} = \text{HR} \times \text{SV} \]

The elevated HR, caused by decreased parasympathetic and increased sympathetic activity to the sinoatrial node, is the primary factor contributing to the raised CO. The SV increase is mainly due to enhanced contractility, mediated by the sympathetic nerves to the ventricular myocardium. Since there is a small increase (about 10%) in end diastolic volume (EDV), the Frank-Starling mechanism also contributes to the increased SV. Increased contractility and the Frank-Starling mechanism are illustrated Figure 4 and in Figure 5. In addition to HR and SV, the following processes act to increase CO and overcome the shortened available filling time at high HR by favoring venous return to the heart: (1) increased skeletal muscle pump activity, (2) increased depth and frequency of inspiration, (3) increased venous tone, and (4) decreased peripheral resistance to blood flow through dilated arterioles.
Figure 4: Increased contractility due to sympathetic ventricular stimulation, showing higher SV at a given EDV [2]

Figure 5: The Frank-Starling mechanism, showing that SV increases as a function of EDV [2]

Myocardial oxygen demand (m\(\text{VO}_2\)) may increase up to 6-fold during exercise, and is a function of heart rate, contractility, ventricular tension, and muscle shortening [6]. The latter two factors constitute ventricular work, and may be determined by the area under the left ventricular pressure-volume (P-V) curve. The differences between the P-V
The left ventricular pressure-volume curve, showing greater ventricular work (area under curve), stroke volume, and contractility with exercise.

An approximate breakdown of the extent to which various factors affect the myocardial oxygen demand and supply during exercise is shown in Figure 7. The main contributor to increased demand is the elevated heart rate, accounting for approximately 60%, with contractility and ventricular work each contributing about 20%. The principal mechanism for augmenting the oxygen delivery is by increasing the coronary blood flow (~80%), which predominantly comes from coronary vasodilation, with a small component due to increased perfusion pressure. The secondary contribution to oxygen supply is the increase in the arteriovenous oxygen difference (~20%). In the presence of stenosis, autoregulation can preserve the basal coronary blood flow at rest, but the
coronary flow reserve (ratio of maximum to basal coronary flow) is reduced. As a result, under the condition of stress, the coronary flow reserve is insufficient to overcome the increased coronary resistance and decreased perfusion pressure distal to the stenosis, leading to insufficient coronary blood supply to meet the elevated oxygen demand [6].

Figure 7: Factors which affect the myocardial oxygen demand and supply during exercise [6]

2.3 Importance of Stress Imaging

Since it was first proposed as a diagnostic tool for angina almost 75 years ago [7], treadmill exercise stress testing quickly became, and still remains, an essential tool in the detection and treatment of heart disease. The Bruce Treadmill Test, first published in 1963 [8], is the most commonly used exercise test protocol in the US [9, 10], and has
been shown to have high diagnostic and prognostic value [11, 12]. During this test, the subject performs treadmill exercise with speed and slope both increasing in 3-minute intervals, while continuously being monitored with a 12-lead electrocardiogram (ECG). Exercise ECG alone, however, has limited sensitivity and specificity, and provides little or no information regarding location and extent of disease. This has led to the addition of imaging to the exercise stress test, providing improved sensitivity, specificity, and diagnostic accuracy compared to exercise ECG alone [13, 14]. Stress imaging is also particularly useful when baseline resting ECG is abnormal, limiting the accuracy of exercise ECG.

According to some estimates, over 10 million stress studies are performed in the US each year in conjunction with nuclear SPECT (Single Photon Emission Computed Tomography) or echocardiographic imaging [15]. SPECT involves the injection of a radioisotope into a patient, which is taken up by cardiac myocytes and distributes within the intracellular space. Gamma ray emission from radioisotopes such as thallium-201 or technetium-99m sestamibi is then detected by gamma ray cameras. In regions of diminished perfusion resulting from coronary stenosis, less radioisotope is taken up by the myocytes, leading to visible perfusion defects in the SPECT images manifested as regions of lower counts. In the case of thallium-201, the radioisotope is injected at peak stress followed by stress imaging 15-30 minutes post-exercise and resting imaging 2.5-4 hours post-exercise in order to evaluate for the presence of reversible, partially reversible, or irreversible perfusion defects [16]. The reason for the 2.5-4 hour delay is to allow time for the radioisotope to redistribute within the myocardium. The long thallium-201 half-
The life of 73 hours limits the radioisotope dose, resulting in relatively low-count density of the images. As a result, technetium-99m is used more commonly due to its shorter half-life of 6 hours, enabling a higher radioisotope dose and improved image quality with less attenuation. Technetium-99m protocols typically start with resting nuclear imaging prior to exercise. The resting radioisotope dose is lower than the stress dose because the quality of the stress study is more critical [16]. One minute before the end of exercise, the radioisotope is injected again, followed by image acquisition at 15 to 60 minutes after exercise [17]. Since technetium-99m is extracted via the hepatobiliary system and excreted into the gastrointestinal (GI) tract, this time delay is required for the radioisotope to clear out of the GI tract in order to avoid subdiaphragmatic attenuation artifact. Optimal timing between injection and imaging has not been extensively studied, and typically depends on liver and GI activity as well as camera availability [17]. Although technetium-99m protocols can be performed over two separate days for rest and stress, it has been shown that comparable diagnostic accuracy can be achieved with one-day rest-stress protocols [18]. Another method to perform one-day protocols is to inject different radioisotopes at rest and stress, relying on the different photon energies of thallium-201 and technetium-99m. However, the radiation dose the patient is subjected to is significantly higher with dual-isotope protocols.

Echocardiography uses the internal tissue reflection of ultrasound waves to generate cine loops of cardiac wall motion. Images at rest are compared to those obtained during cycling or immediately after treadmill exercise to evaluate for the presence of myocardial hypokinesis, akinesis, or dyskinesis not evident at rest. Therefore,
echocardiography aims to detect cardiac wall motion abnormalities in regions of ischemic coronary territories. A meta-analysis of studies comparing stress testing with coronary angiography published between 1990 and 1997 indicates that exercise echocardiography and SPECT both performed better than ECG in detection of CAD [19], justifying the addition of imaging to the exercise ECG test.

2.4 Limitations of Echocardiography and Nuclear Scintigraphy

While stress echocardiography and SPECT are routinely used in the detection of CAD, both are associated with several drawbacks and limitations.

Stress echocardiography has several known limitations which can reduce image quality and limit diagnostic accuracy in some patients. These include poor acoustic windows due to large body habitus, prior cardiothoracic surgery, or lung disease, and poor visualization of the apex of the left ventricular posterolateral wall [20]. Stress echocardiography is highly dependent on the skill of the operator to rapidly acquire proper cardiac views before stress-induced wall motion abnormalities resolve [20]. In addition, poor signal-to-noise ratio and tissue contrast lead to high inter-observer variability for interpretation of the images [20-22].

SPECT also suffers from several known drawbacks and limitations. SPECT involves considerable radiation exposure of between 10 and 28 millisieverts [23]. This radiation dose is in the range in which harmful effects are possible and must be
considered in a patient population likely to undergo other procedures and tests involving significant radiation dosage. SPECT images are of relatively low spatial resolution (on the order of 6-10 mm), and frequently degraded by photon scatter and attenuation artifacts [24], especially in obese patients and women.

In a 2001 comparison based on published data, the sensitivity and specificity of stress echocardiography wall motion (76% and 88%) and stress SPECT perfusion (88% and 77%) to detect CAD were complementary, with echocardiography showing higher specificity, and SPECT higher sensitivity [13]. In addition to image quality, the limited sensitivity of echocardiography can be related to the fact that wall motion abnormalities occur relatively late in the ischemic cascade. SPECT images perfusion defects, which occur earlier in the ischemic cascade, resulting in improved sensitivity over echocardiography. However, the limited specificity of SPECT is primarily due to the photon scatter and attenuation artifact and poor spatial resolution.

Cardiac MRI (CMR) has the potential to overcome these limitations, while providing a more accurate, comprehensive exam, as will be discussed in the following section.

2.5 Potential Advantages of Exercise CMR

CMR is a rapidly developing imaging modality, which has arguably become the “gold standard” for quantitative evaluation of cardiac function [25-27] and myocardial
viability [28-31]. Stress first-pass perfusion CMR imaging followed by delayed enhancement viability imaging has been demonstrated as an accurate method to depict significant coronary stenoses in patients with known or suspected CAD [32].

CMR achieves higher resolution than SPECT, which allows it to discriminate between subendocardial and subepicardial flow. This key advantage gives CMR the capability to non-invasively detect subendocardial perfusion defects, which is the earliest documented event in the ischemic cascade [33, 34] (Figure 8). Example stress SPECT and stress CMR perfusion images are shown in Figure 9. High spatial resolution also allows CMR to avoid the dependency of SPECT on regional hypoperfusion, making CMR more sensitive to globally diffuse subendocardial ischemia [35]. As a result, the specificity and sensitivity of CMR perfusion imaging is at least as good as published data on stress SPECT [36-39]. For example, Ishida et al. [39] have shown that stress perfusion CMR correlates more closely with coronary angiography than does stress SPECT. Table 1 shows the sensitivity and specificity of adenosine stress CMR using PET or coronary angiography as gold standards (Watkins et al. [38], with additional data added from Cury et al. [32], Klem et al. [15], and Al-Saadi et al. [40]). Table 2 shows the sensitivity and specificity of dobutamine stress CMR with coronary angiography as the gold standard (Mandapaka et al. [20], with data added from Schalla et al. [41]). Furthermore, stress CMR has a shorter examination time of 30-60 min compared to 2-3 hours with stress SPECT [42]. This delay with SPECT is due to the time required for the radioisotope to clear out of the myocardium between rest and stress studies with technetium-99m sestamibi, or the time for thallium-201 to redistribute between stress and rest imaging.
Figure 8: Ischemic cascade [34]

Figure 9: Example stress SPECT (left) and stress CMR (right) images. Myocardial perfusion defect (arrow) is clearly seen in CMR, while low spatial resolution of SPECT makes interpretation difficult.
Table 1: The sensitivity and specificity of adenosine stress CMR using PET or coronary angiography as gold standards

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Patients</th>
<th>Sensitivity</th>
<th>Specificity</th>
<th>Comparison</th>
</tr>
</thead>
<tbody>
<tr>
<td>Schwitter J</td>
<td>2001</td>
<td>66 (18 vol)</td>
<td>91/87</td>
<td>94/85</td>
<td>PET/QCA(&gt;50%)</td>
</tr>
<tr>
<td>Ibrahim T</td>
<td>2002</td>
<td>59 (34 vol)</td>
<td>69/86</td>
<td>89/84</td>
<td>QCA(&gt;75%)/PET</td>
</tr>
<tr>
<td>Nagel E</td>
<td>2003</td>
<td>84</td>
<td>88</td>
<td>90</td>
<td>QCA(&gt;75%)</td>
</tr>
<tr>
<td>Ishida N</td>
<td>2003</td>
<td>104</td>
<td>90</td>
<td>85</td>
<td>QCA(&gt;70%)+SPECT</td>
</tr>
<tr>
<td>Plein S</td>
<td>2004</td>
<td>68</td>
<td>88</td>
<td>83</td>
<td>Angiography(&gt;70%)</td>
</tr>
<tr>
<td>Paetsch I</td>
<td>2004</td>
<td>79</td>
<td>91</td>
<td>62</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Takase B</td>
<td>2004</td>
<td>102</td>
<td>93</td>
<td>85</td>
<td>Angiography(&gt;50%)</td>
</tr>
<tr>
<td>Giang TH</td>
<td>2004</td>
<td>80</td>
<td>94/91/94</td>
<td>25/78/71</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Plein S</td>
<td>2005</td>
<td>102 (10 vol)</td>
<td>88</td>
<td>82</td>
<td>Angiography(&gt;70%)</td>
</tr>
<tr>
<td>Curry, RC</td>
<td>2006</td>
<td>46</td>
<td>87</td>
<td>89</td>
<td>QCA(&gt;70%)</td>
</tr>
<tr>
<td>Klem, I</td>
<td>2006</td>
<td>100</td>
<td>89</td>
<td>87</td>
<td>QCA(&gt;70%)</td>
</tr>
<tr>
<td>Al-Saadi, N</td>
<td>1999</td>
<td>34</td>
<td>90</td>
<td>83</td>
<td>QCA(&gt;75%)</td>
</tr>
</tbody>
</table>

Table 2: The sensitivity and specificity of dobutamine stress CMR with coronary angiography as the gold standard

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Patients</th>
<th>Sensitivity</th>
<th>Specificity</th>
<th>Comparison</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensky</td>
<td>1998</td>
<td>6</td>
<td>87</td>
<td>87</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Van Rugge</td>
<td>1994</td>
<td>39</td>
<td>91</td>
<td>80</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Nagel</td>
<td>1999</td>
<td>208</td>
<td>86</td>
<td>86</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Hundley</td>
<td>1997</td>
<td>153</td>
<td>83</td>
<td>83</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Paetsch</td>
<td>2004</td>
<td>79</td>
<td>89</td>
<td>80</td>
<td>QCA(&gt;50%)</td>
</tr>
<tr>
<td>Schalla</td>
<td>2002</td>
<td>22</td>
<td>88</td>
<td>83</td>
<td>QCA(&gt;75%)</td>
</tr>
</tbody>
</table>

Unlike echocardiography, CMR requires no “acoustic window” and can freely visualize any plane in virtually any patient. Using CMR, standard cardiac views can be prescribed at rest, and exactly reproduced at stress without further operator interaction. In addition, CMR provides greater tissue contrast and signal-to-noise than echocardiography (Figure 10). Nagel et al. [43] showed significantly higher diagnostic accuracy for dobutamine stress CMR compared to dobutamine echocardiography in patients with
suspected CAD, and Hundley et al. [44] showed the utility of dobutamine CMR in patients not well suited for dobutamine echocardiography.

Figure 10: Echocardiography (left) and CMR (right) images of the heart in the short axis view. CMR shows greater tissue contrast and signal-to-noise.

In addition to these advantages, CMR is a comprehensive imaging modality that can provide both wall motion and perfusion information at peak stress in addition to viability information obtained at rest. This leads to the expectation that improved diagnostic accuracy can be achieved with a comprehensive exercise CMR protocol including wall motion, perfusion, and viability, with improved resolution and image quality over echocardiography and SPECT.

2.6 Advantages of Exercise over Pharmacologic Stress

Exercise is preferred to pharmacologic testing because it links physical activity to symptoms and ischemia [45]. The exercise test itself offers additional important
information such as exercise capacity, blood pressure response, development of arrhythmias, and the presence or absence of symptoms such as chest pain during exercise [46]. Certain exercise parameters alone such as the inability to complete 6 minutes of the Bruce treadmill protocol [47] and inability to reach 85% of maximum age-predicted heart-rate (MPHR = 220 – age) indicate significant risk of coronary events [48]. This additional information derived from exercise adds to the diagnostic and prognostic value of a stress imaging test.

Besides the reduced information available from pharmacologic stress compared with exercise testing, side-effects are another potential disadvantage. Dobutamine is an adrenergic agent that increases myocardial contractility, heart rate, and blood pressure, with complications including nausea, headache, atypical chest pain, atrial and ventricular arrhythmias, and hyper- or hypotension [49]. Hypotension during dobutamine is not necessarily a manifestation of severe ischemia, as it is in exercise. Vasodilators adenosine and dipyridamole are used to assess coronary perfusion. These agents cause maximal vasodilation in normal epicardial coronary vessels, while stenotic vessels do not increase flow normally. Side effects include chest pain, headache, nausea, dyspnea, and atrioventricular block [50].

Pharmacologic stress is only indicated in those patients unable to undergo exercise stress testing due to de-conditioning, peripheral vascular disease, and orthopedic disabilities [51, 52]. Despite these limitations and drawbacks, the lack of MRI-compatible
exercise and monitoring equipment has made pharmacologic stress the only practical MRI approach to cardiovascular stress imaging to date.

2.7 Current State of the Art in Exercise CMR Devices

The two types of exercise devices used almost exclusively in exercise stress testing are the treadmill and bicycle ergometer. Attempts at exercise CMR using both devices have had limited success due to technical and physiologic challenges. Previous studies of exercise CMR are summarized here.

2.7.1 Bicycle Ergometry

Bicycle ergometry appears to be the natural choice for exercise CMR because it is “patient-powered” and does not require an electromagnetic motor. However, it has significant limitations.

A supine bicycle ergometer that allows imaging during exercise inside a closed-bore magnet is available from Lode BV (Groningen, Netherlands), and is shown in Figure 11. However, pedaling in a totally supine position is uncomfortable and exercise time is limited by the onset of leg fatigue. Depending on the height of the patient and the size of the magnet bore, there may be insufficient knee-to-bore clearance while cycling. This supine ergometer has been primarily used in research studies that did not require maximal
exercise [53] [54]. For example, Niezen et al. [54] performed measurements of aortic and pulmonary flow at two levels of submaximal exercise in 16 healthy volunteers. Even within this group of healthy subjects exercising at relatively low workloads, one subject could not complete the protocol because of muscle fatigue. This problem would be compounded in de-conditioned patients with known or suspected heart disease.

Figure 11: Supine bicycle ergometer (Lode BV, The Netherlands) positioned on the MRI patient table

A more comfortable alternative is the open-magnet bicycle ergometer, where the subject is seated upright inside a vertical open magnet bore and pedals a magnet-compatible bicycle. Similar to the supine ergometer, imaging can be performed during exercise, but the upright cycling position enables higher levels of exercise in a more natural and comfortable exercise posture [55]. The primary disadvantage of this strategy is it requires the use of a vertical, open-bore magnet. While such MRI systems are now
commercially available (Fonar Corp., Melville, NY), this magnet configuration is not in widespread use and has several distinct disadvantages over conventional closed bore magnets. The low magnetic field strength of 0.6T, compared to 1.5 T or 3 T, results in a low signal-to-noise ratio and reduced capacity for real-time cardiac imaging. It is more difficult to construct high amplitude, fast-switching gradient coils for open bore magnets, also adversely affecting cardiac imaging performance. The magnet bore gap is only 46 cm, compared to a conventional 60 cm or 70 cm bore, precluding its use in large patients. Finally, diagnostic 12-lead ECG monitoring is not possible during exercise inside the magnet due to magneto-hydrodynamic effects [56]. This is unsafe and does not meet the American Heart Association guidelines for maximal exercise testing [51, 57].

Figure 12: Open-magnet bicycle ergometer which allows cycling in a seated position [58]
2.7.2 Treadmill

Although bicycle ergometry appears suitable for bloodflow studies during submaximal exercise, treadmill exercise is preferred for cardiovascular stress testing in the United States. The Bruce Treadmill Test, first published in 1963 [8], is the most commonly used exercise test protocol in the US [9, 10], and has been shown to have high diagnostic and prognostic value [11, 12]. Treadmill exercise is subject to less fatigue of the quadriceps muscles than cycling, and untrained subjects will typically only achieve 80% - 90% of their treadmill O$_2$max on a bicycle ergometer [59]. In addition, treadmill does not require active subject cooperation to maintain a specific workload level, which is a significant benefit with de-conditioned cardiac patients. It should be noted that although treadmill is the preferred method of exercise in the US, supine bicycle exercise is preferred in some European countries. The argument for supine exercise is increased venous return, resulting in greater EDV and greater SV, as well as higher systolic and diastolic blood pressure [60]. The combination of higher volumes and pressures leads to increased myocardial wall tension and consequently the myocardial oxygen demand [60]. In addition, the European population tends to be more accustomed to cycling than the US population. Therefore, the potentiating effect of supine exercise on ischemia, the ability to image during peak exercise instead of post-exercise, and the cultural preference for cycling are the likely reasons why supine bicycle may be more common in Europe despite the lower peak heart rate and workload achieved. The protocols used in supine cycling typically increment the workload by 25 W every 2 or 3 minutes [51].
2.8 Exercise CMR Requirements and Challenges

This section will focus on the challenges of treadmill use within the MR environment and the factors that necessitate treadmill placement immediately adjacent to the MR table, followed by a discussion of the imaging requirements imposed by the high heart rates and heavy breathing associated with exercise, and the real-time and single-shot imaging techniques used to overcome these challenges.

2.8.1 Treadmill Challenges in MR Environment

While upright treadmill exercise is the physiologically preferred method of cardiovascular stress testing, it presents significant challenges for use with MRI. Treadmills are typically powered by electromagnetic motors and contain a multitude of ferromagnetic parts, precluding their use in close proximity to an MRI magnet. Rerkpattanapipat et al. [61] have shown the feasibility of detecting severe coronary artery stenoses by exercise CMR using a treadmill positioned outside the magnet room. Safe positioning of the exercise and monitoring equipment required the patient to walk about 20 feet from the treadmill to the MRI system. However, even with the use of segmented imaging sequences requiring breath-holding, they showed that multi-slice cine images of cardiac function could be completed within 60 – 90 seconds post-exercise. The sensitivity and specificity to detect >70% coronary artery diameter narrowing in 27 patients were 79% and 85%, respectively. While the results of this small study are encouraging, a
totally MRI-safe treadmill that can be positioned directly adjacent to the MRI scan table, enabling the MRI scanner room to be configured exactly like a stress-echocardiography lab (Figure 13), is essential for exercise CMR to be safe and successful in typical patients requiring cardiac stress testing, for the reasons outlined below.

Figure 13: Configuration of a stress echocardiography lab

It is essential to acquire images as quickly as possible post-exercise in order to capture transient exercise-induced wall motion abnormalities (WMA) that can rapidly resolve after ischemia is reversed [62-69]. The persistence of WMA’s is most likely related to the severity of CAD (number of vessels involved, percent stenosis), the presence of coronary collateral flow, and the duration of ischemia [70-73]. Therefore, in order to accurately diagnose patients with less severe single-vessel CAD, associated with a high ischemic threshold and rapid WMA resolution, imaging must be performed as close as possible to peak exercise, and ideally with no delay. Dagianti et al. [66] observed that 26% of WMA’s disappeared within 1 minute of the end of supine bicycle exercise. Peteiro et al. [74] demonstrated that multi-vessel CAD would have been
misdiagnosed in 24% of their patients if echocardiographic imaging had been done only post-exercise (49±15 seconds) as opposed to at peak exercise. Presti et al. [65] showed a greater sensitivity for imaging at peak exercise than post-exercise (100% vs. 70%), as did Ryan et al. [75] (91% vs. 83%), Hecht et al. [67] (94% vs. 83%), and Dymond et al. [64] (100% vs. 78%). In the Presti study, of the 29 patients who had a new WMA detected with exercise echocardiography, six would have been misclassified as normal if only imaging done after exercise had been performed. Dagianti et al. [76] found that the duration of transient WMA’s in patients who have undergone percutaneous transluminal coronary angioplasty was only 47±19 seconds. Nakashiki et al [77] developed an ultrasound transducer for continuous echocardiographic imaging during treadmill exercise, and found a greater number of abnormally contracting LV segments at peak than post-exercise. Thus, every effort must be made to minimize the time between end of exercise and imaging in order to maximize the sensitivity of the test, necessitating treadmill placement beside the MRI table.

One potential solution is a partially ferromagnetic treadmill retaining its original electromagnetic motor, and positioned in the corner of an MRI room where the field is less than 5 Gauss. A field less than 5 Gauss poses little or no risk to attracting ferromagnetic objects [78]. However, this poses a significant risk of human error in moving the treadmill too close to the magnet, where it can become a projectile. Permanently anchoring the treadmill in the corner of the room is not possible due to obstruction of patient entry and exit points. This results in having to move the treadmill in and out of the MRI room for each exercise test, which is physically demanding and
highly impractical. In addition, this is not a viable solution at higher fields such as 3 Tesla. Therefore, for exercise CMR to become clinically practical, the treadmill must be completely non-ferromagnetic.

In addition, requiring the patient to walk from a treadmill positioned any distance from the MRI table, whether inside or outside the room, would create a potential safety concern. Immediately following maximal exercise patients may become lightheaded and subject to falling. Only a totally MRI-compatible treadmill would allow for the patient to go directly from the treadmill to the scanner table, as in exercise stress echocardiography.

2.8.2 Imaging Requirements and Challenges

Imaging immediately post-exercise, when the patient is likely to be short of breath and unable to breath-hold, requires real-time and single-shot imaging techniques which significantly shorten scan time and eliminate the need for patient breath-hold. Temporal resolution requirements are even higher for stress imaging than resting cine due to elevated heart rates, and potentially rapid, heavy breathing. Setser et al. [79] have quantified the temporal resolution requirements needed for accurate quantification of left ventricular volumes. Figure 14 shows the minimum sampling rate as a function of heart rate and the number of frequency harmonics based on the Fourier Transform of the subject’s left ventricular volume-time curve. The minimum sampling rate is expressed as $2 \times n \times HR / 60Hz$, where $n$ is the number of frequency harmonics and $n \times HR$ represents
the highest frequency component that will be acquired. Figure 15 presents the mean error, plus 1 standard deviation (SD) above the mean, as a function of the number of frequency harmonics. This plot displays diminishing error with higher harmonics, and shows that the mean error plus 1 SD is within 13% when 4 harmonics are used, within 11% when 6 harmonics are used, and within 10% when 8 harmonics are used. At the start of stress imaging, the heart rate is expected to be approximately 85% of MPHR, such that for an 18-year old subject, the expected heart rate would be 0.85 x (220 -18) = 172 bpm, resulting in a temporal resolution requirement of 43.6 msec when using 4 harmonics and 21.8 ms when using 8 harmonics. This requirement is lower for older subjects, as shown in Figure 16 for 4 harmonics and Figure 17 for 8 harmonics. This corresponds well with the known timing of events in the cardiac cycle; the isovolumetric relaxation (end systole) is 70-80 msec over a wide range of ages, and isovolumetric contraction (end diastole) is 38±10 msec in men and 39±9 msec in women [79]. In addition, the slow rate of change in LV volume immediately preceding and following isovolumetric contraction is likely to reduce the stringency of the sampling requirements [79]. Since the principal parameter that changes during stress is the shortened filling time, these temporal resolution requirements are not expected to vary appreciably with stress.
Figure 14: The minimum sampling rate required to accurately quantify left ventricular volumes as a function of heart rate and of the number of frequency harmonics obtained from the Fourier transform of the left ventricular volume-time curve [79]

Figure 15: The error in left ventricular volumes as a function of the number of frequency harmonics, displaying decreasing error with higher harmonics [79]
Figure 16: The temporal resolution required to quantify left ventricular volumes as a function of age using 4 harmonics [79], showing that a lower temporal resolution is required with increasing age.

Figure 17: The temporal resolution required to quantify left ventricular volumes as a function of age using 8 harmonics [79], showing that a lower temporal resolution is required with increasing age.
The principles of real-time cine and single-shot perfusion imaging will be discussed in the following section.

2.8.2.1 Cine Imaging

The cine pulse sequence based on balanced Steady-State Free Precession (SSFP) is illustrated in Figure 18 [80] which shows no net dephasing between RF pulses. This sequence has high signal-to-noise due to short echo time, and good blood/myocardium contrast. This type of sequence was originally applied in dynamic cardiac imaging using segmented algorithms, which acquire a segment of k-space corresponding to a cardiac phase during a single heartbeat, and progressively build up the full k-space over multiple heartbeats. Although this method has high temporal resolution, it requires breath-holding and ECG triggering, and is impractical for peak exercise. The advent of parallel imaging techniques such as SENSE [81] and GRAPPA [82] have enabled us to under-sample k-space in each dynamic frame by a factor known as the acceleration rate, thus speeding up data acquisition sufficiently to enable real-time imaging. For example, acceleration rate 2 with every second line acquired is depicted in Figure 19, leading to the aliased image shown on the right. The missing information required to obtain the final un-aliased image shown in Figure 20 is obtained by computing the spatial coil sensitivity maps, using the fact that the coil sensitivity decreases as a function of distance from the coil. The coil sensitivity maps in SENSE and GRAPPA are obtained either during a pre-scan, leading to a penalty in acquisition time and poor accuracy in the case of free breathing, or by
acquiring extra lines in the center of k-space with each frame. While the advantage of the latter approach is that the coil sensitivity maps are updated with each frame, this comes with a penalty in temporal resolution due to the additional acquired k-space lines.

![Balanced Steady State Free Precession (SSFP) sequence diagram](image)

**Figure 18**: Balanced Steady State Free Precession (SSFP) sequence, showing no net dephasing between the RF pulses ($G_{ss}$ = slice select gradient, $G_{PE}$ = phase encode gradient, $G_{R}$ = readout gradient, $T_s$ = sampling time) [80]

![Rate 2 parallel acceleration diagram](image)

**Figure 19**: Rate 2 parallel acceleration with every second line of k-space acquired, resulting in aliasing as shown in the example on the right (a: acquired, s: skipped)
Figure 20: Un-aliased version of the image shown in the previous figure. Parallel imaging was used to provide additional information from the spatial coil sensitivity maps in order to reconstruct the full FOV image.

More recent advances, such as the TSENSE [83] and TGRAPPA [84] methods of dynamic parallel imaging, have resulted in further improvements over SENSE and GRAPPA in the way the coil sensitivity maps are obtained. These methods interleave under-sampled dynamic frames in order to reconstruct the full-resolution coil sensitivity maps. No additional lines are acquired in the center of k-space, leading to improved temporal resolution. Figure 21 shows the example of rate 2 acceleration, where frames 1 and 2, and frames 2 and 3 are combined into full-resolution coil-sensitivity maps, which may be averaged to improve signal-to-noise.
Figure 21: An example of interleaving the under-sampled dynamic frames in order to reconstruct full resolution coil sensitivity maps at rate 2 parallel acceleration.

Additional improvement in data acquisition may be obtained with a 32-element cardiac array coil instead of a standard 12-element array. Signal-to-noise ratio is known to decrease with acceleration rate $R$ due to fewer acquired k-space lines and the spatially-varying coil geometry factor ($g$) as follows [85]:

$$SNR_{accelerated} = \frac{SNR_{unaccelerated}}{\sqrt{R / g(x, y, rate)}}$$

Wintersperger et al. [86] have demonstrated that acceleration rate of 4 in resting cardiac cine imaging is feasible with a 32-channel array coil, which yields a lower $g$-factor than the standard 12-element array due to the larger number of coil elements. The application of parallel imaging methods with the 32-channel receiver array can result in real-time acquisition with improved spatial resolution, temporal resolution, and/or signal-to-noise [87]. Therefore, the benefit of the 32-channel array for high heart-rate, free breathing imaging may be gains in temporal resolution at higher acceleration rates, or providing
higher quality images with less noise and artifact at standard acceleration rates. The use of a 32-channel array coil in exercise stress testing will be described in Section 4.2.

2.8.2.2 Perfusion Imaging

Figure 22 illustrates the principle of CMR perfusion combined with delayed enhancement viability imaging. On the first pass of the contrast agent through the heart, the ischemic myocardium is less enhanced than the normally perfused regions since there is reduced blood flow and hence less contrast agent delivered through stenotic coronary arteries. While the extracellular contrast agent cannot penetrate the viable cell membrane, the cell membrane integrity is compromised in non-viable cells. In addition, the infarced regions which have been replaced by fibrous tissue have a sufficiently low density to allow contrast diffusion. As a result, the extracellular contrast agent diffuses through the infarcted and fibrotic tissue, resulting in hyper-enhancement of non-viable myocardium five or more minutes post-injection.
Figure 22: The principle of first-pass perfusion and delayed enhancement, which are used to differentiate between normal, ischemic, and infarcted myocardium

In addition to standard visual assessment of first-pass perfusion, a potentially useful quantitative perfusion indicator is the myocardial perfusion reserve index (MPRI), which is the ratio of the slope of the myocardial signal intensity curve at stress and rest, normalized by the slope of the left ventricular (LV) blood pool signal intensity curve. The normalization is intended to compensate for the increased CO at stress. Figure 23 shows typical signal intensity curves for a myocardial segment at stress and rest, with a normal segment shown on top and a hypoperfused segment on the bottom [5].
Figure 23: Myocardial signal intensity curves following the injection of contrast agent. A normal myocardial segment is shown on top, illustrating the increased slope and the higher peak intensity at stress. The hypoperfused segment on the bottom shows no signal increase at stress. [5]

A single-shot first-pass perfusion pulse sequence using the Gradient Recalled Echo – Echo Planar Imaging (GRE-EPI) is illustrated in Figure 24. Figure 25 shows the magnetization recovery following a 90° saturation recovery pulse. The magnetization in the non-ischemic myocardium recovers faster than the ischemic myocardium due to the greater concentration of T1-shortening contrast agent in the normally perfused regions,
and data is collected during an acquisition period after sufficient contrast is achieved between the ischemic and non-ischemic myocardium. Multiple slices are acquired within a single heartbeat, as shown in Figure 26, and measurements typically extend over 50-60 heartbeats. To the author’s knowledge, Jekic et al. [88] have been the first to demonstrate exercise stress first-pass perfusion CMR.

Figure 24: First-pass perfusion pulse sequence using the Gradient Recalled Echo – Echo Planar Imaging (GRE-EPI) [89]
Figure 25: Magnetization recovery following a $90^\circ$ saturation pulse, showing a greater recovery in the normal compared to the ischemic myocardium due to the presence of T1-shortening contrast agent (TR/TE 5.8/1.2 ms, matrix 96x160, rate 2 parallel acceleration, echo train length 4, saturation period 30 ms, acquisition period 70 ms, normal T1 100 ms, ischemic T1 400 ms)
Figure 26: The acquisition of multiple slices within a single heartbeat using GRE-EPI. In this example, 3 slices are acquired using separate 90° saturation pulses.
CHAPTER 3

FEASIBILITY OF EXERCISE CMR IN HEALTHY SUBJECTS

This chapter will describe a concept feasibility study for exercise CMR by applying minimal modifications to existing equipment and imaging methods. The aim of this study was to demonstrate the feasibility of exercise CMR in 20 healthy subjects, including stress wall motion and stress myocardial perfusion, using a partially MRI-compatible treadmill located in a low magnetic field region in the corner of the MRI room. The heart rate increase, ejection fraction increase, and the cardiac output increase with exercise were quantified based on the cine wall motion data, and the myocardial perfusion reserve index was quantified based on the perfusion data. The technical developments required to make exercise CMR fully clinically realizable were identified based on this feasibility study and will be described in subsequent chapters.

3.1 Subjects

Twenty healthy subjects between the ages of 20 and 64 (11 females, 9 males, mean age 39±13 years, median age 44) underwent the exercise CMR protocol. The study protocol was approved by the Institutional Review Board at The Ohio State University.
All participants gave written informed consent. The exclusion criteria were known or suspected CAD, inability to exercise, and standard contraindications to MRI.

3.2 Equipment

The requirements for performing exercise CMR inside the MRI room include the following:

1. A treadmill inside the MRI room
2. Continuous 12-lead ECG and blood pressure monitoring during exercise and recovery
3. Positioning the patient at the same location on the MRI table for rest and stress imaging
4. Starting the stress scan from inside the MRI room in order to commence imaging as quickly as possible post-exercise and remain in communication with the patient at all times

The methods by which the above challenges were addressed are described below.

Treadmill modifications

We modified a standard rehabilitation treadmill (Landis 8700, Randolph, NJ) by replacing the ferromagnetic components to the rear of the motor with non-ferromagnetic stainless steel and aluminum equivalents, as shown in Figure 27. Replacing these parts
allowed us to safely position the treadmill in the corner of the MRI room such that the remaining ferromagnetic components, including the motor, front roller, and elevation mechanism, were located in a magnetic field of approximately 0.0002 Tesla (2 Gauss). A magnetic field of less than 5 Gauss poses little or no risk of attracting ferromagnetic objects [78]. Figure 28 shows the magnetic field plot, which was verified with a hand-held Gaussmeter, and the room layout for the MRI system. The outline of the treadmill positioned in the corner of the MRI room is shown, along with a shaded area indicating the location of the remaining ferromagnetic parts.

Figure 27: Treadmill modified by replacing ferromagnetic components to the rear of the motor assembly with non-ferromagnetic stainless steel and aluminum
Patient monitoring

Continuous 12-Lead ECG monitoring of the patient is required during the exercise test [51]. To our knowledge there is currently no commercially available, MRI-compatible 12-lead stress ECG system. Therefore, we positioned a standard 12-lead ECG system at the entrance to the MRI room (Figure 29), close enough to monitor the subject both on the treadmill and on the MRI patient table when the patient is outside of the magnet bore. While inside the bore, the ECG is non-diagnostic due to magneto-hydrodynamic artifacts caused by blood flow within the magnetic field [90]. However, heart rate and rhythm can be monitored continuously with a wireless 3-electrode unit.
provided by the MRI manufacturer (Siemens Medical Solutions, Malvern, PA). This setup allowed us to quickly disconnect the patient from the 12-lead ECG system after exercise, while continuing to monitor heart rate with the 3-electrode unit. MRI-compatible manual and automatic non-invasive blood pressure equipment (Medrad, Inc., Pittsburgh, PA) was used to monitor blood pressure before, during, and after the stress test.

![Image of experimental setup](image)

Figure 29: Experimental setup for the treadmill CMR test inside the MRI room. The ferromagnetic components, including the treadmill motor and the ECG system, are located in the corner of the MRI room where the magnetic field is less than 5 Gauss.

**Patient positioning**

Before exercise, the subject was positioned on the MRI table using two vacuum mattresses (Vac-Lok Cushions, MEDTEC, Orange City, IA), and slice localization and
resting function scans were performed. One vacuum mattress was placed under the head and shoulders and the other under the legs extending from foot to upper thigh (Figure 30). Removal of air with a vacuum pump causes the mattresses to rigidly conform to the body. These devices are commonly used for repositioning of patients undergoing repeated radiation therapy sessions. This system was used to ensure that the subject returned to the same position after exercise such that rest and stress imaging could be performed at the same position.

Figure 30: Two vacuum mattresses (head and shoulders, and foot to upper thigh) are used for repositioning the subject between rest and stress. Removal of air with a vacuum pump causes the mattresses to form a rigid mold around the subject.
We used a 1.5 Tesla MAGNETOM Avanto MRI system (Siemens Medical Solutions, Malvern, PA) with maximum gradient strength of 45 mT/m and slew rate of 200 mT/m/sec, 32-channel RF system, and a 12-channel array coil with 6 anterior and 6 posterior elements. The stress MRI scans were started using a start button located on the front panel of the magnet housing.

A power injector (Spectris, Medrad Corp., Pittsburgh, PA) was outfitted with a manual control switch for operation from within the MRI room. The injection protocol was pre-programmed and loaded such that it could be executed immediately at the start of the perfusion scan from within the MRI room.

Thus, all equipment necessary to conduct the treadmill exercise test, including a partially MRI-compatible treadmill, continuous ECG and blood pressure monitoring, as well as the injection of contrast agent during stress perfusion, was positioned to allow the test to be performed within the MRI room. The stress testing team was able to remain in the room and in direct communication with the patient at all times.

3.2.1 Equipment Update

Since this feasibility study was performed, several equipment updates have taken place, which are illustrated in Figure 31. Although the equipment described in the
previous section was used to conduct the studies presented in this chapter, a brief overview of the new setup is provided here for contrast with the original. Instead of the ECG machine positioned in the doorway of the MRI room, Cardiosoft stress ECG software (GE Healthcare, Piscataway, NJ) was installed on a PC in the control room, and the ECG traces are displayed during exercise and recovery using an MRI-compatible in-room monitor (Siemens Medical Solutions, Malvern, PA). The communication between the PC and the keyboard and mouse is achieved via a Bluetooth transceiver located inside the MRI room, which is connected to the PC using a shielded USB cable through a waveguide. In addition, a fully MRI-compatible treadmill was developed which is positioned immediately adjacent to the MRI table. The MRI-compatible treadmill and the equipment communications will be discussed in detail in Chapter 7.

Figure 31: New equipment setup
3.3 Pulse Sequences

A real-time SSFP sequence with TR/TE of 2.3/1.0 msec and TSENSE acceleration factor of 3 was used for cine function imaging. Five slices were acquired in the short axis (SAX) direction, and one slice each in horizontal (HLA) and vertical (VLA) long axis directions. Temporal resolution of $57.8 \pm 3.4$ msec and spatial resolution of $2.9\text{mm} \times 3.7\text{mm} \times 8\text{mm}$ were achieved with no breath-hold and no ECG gating. The seven slices were acquired in an average of 15 seconds.

0.1 mmol/kg gadolinium-DTPA was administered intravenously at a rate of 4 mL/s as a contrast agent for first-pass perfusion imaging. GRE-EPI with TR/TE of 5.8/1.2 msec and TSENSE acceleration rate of 2 was used to obtain 3 SAX slices each cardiac cycle. Saturation recovery time was 30 msec and the acquisition time per slice was 70 msec (96 x 160 matrix, 3.0mm x 2.4mm x 10mm resolution).

3.4 Data Analysis

Argus (Siemens Medical Solutions, Inc., Malvern, PA) was used to compute the ratio of cardiac output (CO) at stress and rest by visually selecting the end diastolic and end systolic frames, and manually tracing the endocardial borders. We selected the 3 out of 5 SAX slices that were most closely aligned between rest and stress in order to compensate for any repositioning error.
The myocardial perfusion reserve index (MPRI) was computed for one mid-ventricular slice using Argus. For these healthy volunteers, uniform myocardial perfusion was assumed in the analysis. The slope of the myocardial signal intensity curve was computed for the entire slice and normalized by the slope of the blood pool signal intensity curve [5]. The stress slope was divided by the rest slope to obtain a global estimate of MPRI.

3.5 Exercise CMR Protocol

The exercise CMR protocol is summarized in Figure 32. Patient preparation included insertion of an intravenous (IV) needle and the standard placement of both the 12-lead and the 3-electrode wireless ECG electrodes on the chest [51]. Supine 12-lead ECG and blood pressure (BP) were recorded at rest. The supine resting ECG was required for direct comparison with the supine recovery ECG post-exercise. Next, subjects were positioned on the MRI table using the vacuum mattresses. Air was removed from the mattresses through a vacuum line located inside the MRI room.
Figure 32: Timeline for the treadmill CMR test, including slice localization, rest and stress function, rest and stress perfusion, and viability. Both the duration of each stage of the test and the estimated cumulative time are shown.

Slice localization by single-shot SSFP was followed by resting cine imaging. The cine sequence was set up to scan each slice position for 1.8 seconds, while the temporal resolution varied depending on the size of the patient and the resulting field of view. A test acquisition for first-pass perfusion was performed without contrast agent. The pulse sequences were queued up for stress imaging such that they could be executed automatically from the scan start button located on the magnet. The subject was then removed from the magnet, with care taken not to pull the table all the way out of the magnet, and not to move the surface array coil too drastically to avoid activation of a Hall-effect sensor in the cardiac array coil. Either of these actions was found to cause the system to repeat adjustments prior to the start of the stress scan, causing a delay and/or failure to execute the cine scan.
Next, the subjects exercised on a treadmill positioned inside the MRI room, as previously shown in Figure 29. The treadmill speed and elevation were progressively increased every three minutes following the standard Bruce protocol (Table 8 and Table 10 in Chapter 7). 12-lead ECG was continuously monitored during exercise. Blood pressure was measured and a hard copy of the ECG was obtained at the midpoint of each Bruce protocol stage. As with conventional stress testing, subjects were continuously monitored by a nurse and/or physician who could stop the test at any time based on recognition of adverse endpoints or in response to the subject’s request. After reaching their exercise limit or the maximum predicted heart rate (MPHR) based on age (220-age), the 12-lead ECG was disconnected and the subjects were quickly escorted to the MRI table. Heart rate and rhythm monitoring continued with the 3-electrode wireless unit. The surface coil was placed on the chest, the contrast injector was connected to the previously inserted IV in the subject’s arm, and the MRI table was returned to the original position inside the magnet. The previously prepared cine and first-pass perfusion scans were started using the start button located on the magnet. Stress function was executed first, followed by stress perfusion. The time from end of exercise to start of imaging (Tstart) was recorded. We started the injection protocol as soon as the audible change from the cine pulse sequence to the first-pass pulse sequence was detected. The subject remained inside the magnet bore for approximately 90 seconds for stress imaging.

Immediately following imaging, the MRI table was pulled out and 12-lead ECG and blood pressure were recorded for 6-8 minutes during supine recovery on the table. Following recovery, the subject was again moved into the magnet bore, and resting
perfusion and late Gadolinium enhancement (LGE) viability images were obtained. We
grew the acquisition order described by Klem et al. [15], which included stress
perfusion followed by rest perfusion and finally viability. Stress perfusion is performed
prior to rest perfusion since the strongest image contrast and the highest sensitivity to
perfusion defects occur with the first injection of Gadolinium contrast agent. Resting first
pass images are primarily used to distinguish image artifacts from true perfusion defects.
This is the standard order of acquisition for vasodilator stress perfusion MRI also [15].

3.6 Healthy Volunteer Results

The entire procedure, from initial setup to completion of the viability scan, took
an average of one hour. The results are displayed in Table 3 for the 19 out of 20 subjects
for whom at least one of the stress scans (function or perfusion) was successful. Two
perfusion scans were unsuccessful due to IV failure, and two function scans failed to start
due to activation of the Hall-effect motion sensor in the array coil, which caused the MRI
system to automatically perform a series of adjustments taking several seconds or skip to
the next scan in the queue when the coil was moved too drastically inside the magnetic
field. The first time the function scan failed, no usable data was obtained, whereas in the
second case perfusion data was acquired. The former case is not presented in Table 3
since no image data was obtained.
Table 3: Results of the treadmill CMR test in 19 healthy subjects. The heart rate at peak exercise and start of imaging is expressed as %MPHR. The times from end of exercise to start of imaging, from end of exercise to end of function imaging, and from end of exercise to peak myocardial enhancement are shown. HR: heart rate, MPHR: maximum predicted heart rate based on age, CO: cardiac output, MPRI: myocardial perfusion reserve index

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<td>2.2</td>
<td>28</td>
<td>40</td>
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| Mean    | 13  | 5           | 98             | 84                   | 3.1       | 1.9     | 30   | 45            | 57               | 57                |
| SD      | 13  | 1           | 7              | 11                   | 0.6       | 0.5     | 4    | 4             | 5                | 5                 |

Heart rate at peak exercise and at start of imaging is expressed as percent of maximum predicted heart rate based on age (MPHR = 220-age). Cardiac output increased by a factor of 3.1±0.6 from rest to stress. Increased contractility is clearly depicted in the end-systolic images at rest and stress displayed in Figure 33. Figure 34 shows representative first-pass perfusion images at stress and rest, and the myocardial signal intensity curves measured during the first pass of the contrast agent through the heart. The increase in slope from rest to stress due to increased flow through the myocardium is
clearly depicted, which supports quantitative perfusion analysis that yielded an average MPRI of 1.9±0.5. The scan commenced an average of 30±4 seconds after exercise while function imaging was completed an average of 45±4 seconds post-exercise. The time from end of exercise to peak myocardial enhancement in the perfusion images was 57±5 seconds. Image quality was sufficient for visual assessment of wall motion in all left ventricular segments, and no subject showed any evidence of exercise-induced wall motion or perfusion abnormality.

Figure 33: End-systolic frames clearly depicting increased contractility from rest (top) to stress (bottom). From left to right, three SAX, one VLA, and one HLA views are shown.
Figure 34: Three short-axis first-pass perfusion images at rest and during exercise stress are shown on the left. Graph on the right shows myocardial signal intensity in a mid-ventricular slice during the first pass of the contrast agent, clearly depicting the increased slope from rest to stress.

3.7 Discussion

We have successfully demonstrated the ability to acquire stress cardiac function and myocardial perfusion images immediately following maximal exercise on a treadmill inside the MRI room in healthy subjects. In the two cases of accidental Hall-effect sensor activation due to significant coil movement within the magnetic field, several seconds of automatic system adjustments were triggered and the cine sequence failed to start. The first time this occurred, contrast was not injected and no usable image data obtained. The second time, we injected contrast at the start of the perfusion scan, enabling us to quantify MPRI in the absence of function data (Subject #14 in Table 1). Once we recognized the cause of this problem, we were able to avoid it in subsequent studies.
The increase in cardiac output by a factor of 3.1 was comparable to a previous study in healthy young sedentary subjects undergoing maximal treadmill exercise, which showed a factor of 3.2 cardiac output increase in men and 2.9 in women [91]. In addition, the MPRI of 1.9±0.5 fell within the normal range of MPRI (2.3±1.3, averaged over six segments) previously found with adenosine CMR in patients with negative coronary angiograms [5]. Since resting perfusion was acquired following recovery from exercise, it was possible that the subjects had not completely recovered and still had elevated blood flow at the time of resting perfusion. This may have artificially lowered the MPRI to some extent. No subject demonstrated chronotropic incompetence, i.e. the inability to reach 85% of MPHR based on age. Except for the oldest subject (age 64), all subjects reached at least stage 4 of the Bruce Protocol (exercise time >9 minutes). The spatial resolution of 2.9 x 3.7 mm was adequate for wall motion assessment and the delineation of endocardial borders for calculating functional parameters.

Rerkpattanapipat et al. [61] have shown the feasibility of detecting coronary artery stenoses by exercise CMR using a treadmill positioned outside the MRI room, using segmented cine pulse sequences requiring a 4 to 8 second breath-hold and no perfusion imaging. The application of real-time imaging for wall motion assessment is an advantage of the current work over that published by Rerkpattanapipat. Real-time imaging methods eliminate the requirements of breath-holding and ECG gating, and provide inherently faster scan time allowing the acquisition of five short-axis and two long-axis views. This enabled sufficient coverage for the assessment of wall motion abnormalities using the standard 17-segment model of the left ventricle [3]. The temporal
resolution of the real-time acquisition in our study was limited to 57.8 milliseconds, and further investigation is required to determine whether this is sufficient to accurately resolve exercise-induced wall motion abnormalities. An additional advancement in our protocol was the addition of stress and rest perfusion and viability information.

While ECG triggering was not required for real-time cine, it was required for perfusion imaging. Triggering difficulties were encountered in some cases due to heavy respiratory chest wall motion post-exercise, but did not prevent the interpretation of the first-pass image data. A mistimed trigger affects the phase of the cardiac cycle at which the image is acquired, but not image quality since we are using a single-shot sequence. The next phase of the feasibility study is to test whether this combined approach of exercise stress with CMR of function, perfusion, and viability will provide improved diagnostic accuracy for CAD over existing methods. In addition, while this work and the work by Rerkpattanapipat et al.[61] provide a critical step toward developing a feasible treadmill exercise CMR system, in order to make treadmill exercise CMR clinically viable, a fully MRI-compatible treadmill that can be positioned next to the MRI patient table is required to minimize the time delay between exercise and imaging, and provide a safe means of CMR exercise stress testing at any MRI field strength.

3.8 Conclusions

This study has successfully demonstrated the feasibility of in-room treadmill exercise CMR in healthy volunteers. However, several technical challenges still need to
be addressed. Imaging optimization is needed to reduce the artifact level and improve the temporal resolution. Further investigation is required to determine whether 12-lead ECG monitoring is accurate within the MRI exam room. Finally, the partially ferromagnetic treadmill in the corner of the MRI room is unsafe and clinically impractical, and will need to be replaced by a fully MRI-compatible treadmill. In addition to these technical challenges to be addressed, confirmation of feasibility in patients with cardiovascular disease is still needed.
The previous chapter showed that an exercise CMR exam combining cardiac function and myocardial perfusion at peak stress with myocardial viability at rest is feasible in healthy subjects. However, further imaging optimization is needed to reduce parallel imaging artifacts and improve the temporal resolution. The first part of this chapter compares the performance of TSENSE and TGRAPPA methods of parallel imaging under the condition of deep breathing encountered in exercise stress testing. The second part of the chapter investigates whether higher parallel imaging acceleration factors can be achieved to improve the temporal resolution by using a 32-channel cardiac array coil.

4.1 Quantitative Comparison of TGRAPPA and TSENSE Parallel Reconstruction

The rapid heart rates and deep breathing encountered during exercise CMR make it difficult to achieve the necessary spatial resolution, temporal resolution, and image
quality with real-time cine techniques. Parallel imaging has enabled improved temporal resolution, but at the cost of increased artifact and noise. Temporal resolution can be further improved by reducing the FOV in the phase encoded direction. Temporal resolution is a function of the number of acquired phase encoded lines (Ny), which can be defined as:

\[ Ny = \frac{\text{FOVy}}{\Delta y} \]  

Equation 1

Therefore, reducing FOVy can improve the temporal resolution without sacrificing the spatial resolution (\(\Delta y\)). However, in the case of deep breathing such as during or after exercise or dobutamine stress, the chest wall can move in and out of the FOV, causing aliasing. Surface coils can also move with deep breathing, causing a mismatch between the coil sensitivity map used in the reconstruction and the actual coil position, which is another potential source of artifact [84].

SENSE is known to create problems when the FOV is smaller than the object [92]. For the example of rate 2 parallel acceleration using N coils, a folded image pixel in coil j (Ij) can be expressed in terms of two aliased pixels (\(\rho_1\) and \(\rho_2\)) scaled by the respective coil sensitivities S as follows [93]:

\[
\begin{align*}
I_1 &= S_{11}\rho_1 + S_{12}\rho_2 \\
I_2 &= S_{21}\rho_1 + S_{22}\rho_2 \\
\vdots \\
I_N &= S_{N1}\rho_1 + S_{N2}\rho_2
\end{align*}
\]  

Equation 2
Solving this system of equations using a pseudo-inverse of the matrix S results in the un-aliased pixels $\rho_1$ and $\rho_2$. When the FOV is smaller than the object, there is signal contribution due to the additional aliasing ($\rho_3$), which is spread out between pixels $\rho_1$ and $\rho_2$ and results in significant residual aliasing artifacts. In addition to aliasing issues, inaccurate coil sensitivity maps can cause the above linear system of equations to become ill-conditioned and can locally amplify noise [94].

GRAPPA, a k-space-based reconstruction method, is less sensitive than SENSE to aliasing and inaccurate coil sensitivity maps [93]. With GRAPPA, each line in k-space is derived through linear combinations of other lines. A reduced FOV results in aliasing since it creates a greater spacing in k-space ($\Delta k = 1/\text{FOV}$) than Nyquist (Figure 35). However, since the signal at each line in k-space is independent of the FOV, spacing to the neighboring lines ($\Delta k$) does not determine signal at any particular line. No k-space lines are thus corrupted by aliasing in the GRAPPA reconstruction [93]. TGRAPPA [84] is an improvement over GRAPPA which does not require additional reference lines to be acquired and thus improves the temporal resolution. Our hypothesis is that TGRAPPA will have better artifact performance than TSENSE under the condition of deep breathing which induces aliasing and coil sensitivity map errors.
While physician-assigned qualitative scoring is typically used to evaluate these artifacts, it is subjective, time-consuming, and impractical for large data sets. To overcome this challenge, we quantified parallel imaging artifacts by exploiting their spatially fixed nature and examining the peak of the autocorrelation function applied to the image series.

4.1.1 Methods

We acquired cine series in three views (short-axis, vertical and horizontal long-axis) in 5 healthy subjects during deep breathing using SSFP real-time cine accelerated 3-fold with TSENSE and TGRAPPA on a Siemens 1.5T Avanto, resulting in a total of 30 image series. Sensitivity maps were obtained by interleaving and averaging all undersampled frames in each series. Scan parameters were: TE/TR 0.9/2.2ms, average

Figure 35: Illustration of the inverse relationship between the field-of-view and the spacing of k-space lines ($\Delta k = 1/\text{FOV}$)
The FOV was specified immediately adjacent to the chest wall in the view in which the body appears largest in the phase encode direction (typically the vertical long axis view). Artifacts were quantitatively analyzed as follows:

1. Subtraction was performed on each adjacent image pair within the series to eliminate most stationary tissue. The spatially fixed artifact is still apparent because it is temporally modulated in rate 3 TSENSE and TGRAPPA such that it fluctuates periodically over every 3 image frames because the k-space lines are acquired at the same position every 3 frames.

![Subtraction of two adjacent frames](image)

Figure 36: Subtraction of two adjacent frames, resulting in the difference image on the right which displays the spatially fixed parallel imaging artifact at 1/3 FOV and 2/3 FOV.
2. The autocorrelation was calculated in the phase encode direction, resulting in a 1D autocorrelation plot for each subtracted image pair.

3. Peaks were measured in the 1D autocorrelation function at shifts of 1/3 and 2/3 FOV, the expected locations for parallel imaging associated ghost artifacts at rate 3 acceleration.

![Autocorrelation Plot](image)

Figure 37: The autocorrelation function in the phase encoding direction, which shows peaks due to the spatially fixed parallel imaging artifact at shifts of 1/3 FOV and 2/3 FOV

4. The highest 10% of the autocorrelation peaks at 1/3 FOV were averaged to define an “artifact index”. We chose to use the highest 10% of autocorrelation coefficients rather than the mean or median under the expectation that only a few inadequate frames would cause the entire image series to receive a low image quality score.

Two experienced readers blinded to the study assigned qualitative scores of artifact severity: (1) none, (2) minor, (3) moderate, and (4) severe, and image quality: (1)
excellent, (2) good, (3) diagnosis may be limited, and (4) non-diagnostic. We compared the TGRAPPA and TSENSE quantitative and qualitative scores by using a Student’s paired T-test, with $p<0.05$ considered statistically significant.

In order to isolate the effect of the coil sensitivity errors from the aliasing effect, we performed measurements in 4 healthy subjects in a transverse slice through the liver to exclude the effects of cardiac motion. In the baseline case, an accurate coil sensitivity map was maintained throughout the scan by breath-holding. In the second case, the coil sensitivity map was acquired at peak inspiration while the remaining data was acquired during end-expiratory breath-holding in order to create a difference between the measured and actual coil sensitivities. We reconstructed each data set using both SENSE and GRAPPA.

4.1.2 Results

In the real-time SSFP cine images, TGRAPPA was superior to TSENSE in terms of the physician-assessed artifact severity and image quality. The results are summarized in Table 4. Figure 38 illustrates the increased artifact level with TSENSE compared to TGRAPPA. The artifact index obtained with the autocorrelation function agreed with the qualitative artifact scores, with TGRAPPA showing less artifact severity. Figure 39 displays the higher autocorrelation coefficient for TSENSE vs. TGRAPPA at 1/3 and 2/3 FOV. All image series with a qualitative artifact score of 1 (best) had an artifact index
<0.075, while all images with an artifact score of 4 (worst) had an artifact index >0.075.

In addition, TSENSE exhibited a greater artifact index in the liver images acquired using inaccurate coil sensitivity maps (Figure 40) both at rate 3 and rate 4 accelerations (p = 0.0025 and 0.0114, respectively).

Table 4: Quantitative comparison between the TSENSE and TGRAPPA methods of parallel imaging reconstruction under the condition of deep breathing

<table>
<thead>
<tr>
<th></th>
<th>TGRAPPA</th>
<th>TSENSE</th>
<th>Best Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Artifact index</td>
<td>0.058±0.035</td>
<td>0.110±0.052</td>
<td>TGRAPPA, p=0.0036</td>
</tr>
<tr>
<td>Artifact severity score</td>
<td>1.7±0.6</td>
<td>2.9±0.8</td>
<td>TGRAPPA, p&lt;0.001</td>
</tr>
<tr>
<td>Overall image quality score</td>
<td>1.6±0.6</td>
<td>2.3±0.7</td>
<td>TGRAPPA, p&lt;0.001</td>
</tr>
</tbody>
</table>

Figure 38: Increased artifact level observed in TSENSE (left) compared with TGRAPPA (right) under deep breathing
Figure 39: Autocorrelation function performed in the phase encode direction on a pair of adjacent image frames subtracted from each other. The figure shows the increased artifact index for TSENSE (TSE) over TGRAPPA (TGR) at parallel acceleration rate 3.

Figure 40: A liver image frame showing an increased artifact level when reconstructed using TSENSE (right) compared to TGRAPPA (left). The artifact results from an inaccurate coil sensitivity map.
4.1.3 Conclusions

We quantified parallel imaging artifacts by exploiting their spatially fixed nature to determine peaks in the autocorrelation function and obtain an “artifact index”. Both quantitative and qualitative scores showed that TGRAPPA delivered superior artifact performance compared to TSENSE in the presence of deep breathing. TGRAPPA also exhibited lower artifact levels due to inaccurate coil sensitivity maps in the absence of aliasing. While SENSE provides an optimal reconstruction under normal conditions of breath-hold or quiet breathing, our results demonstrate that it suffers from artifacts when the FOV is smaller than the object or the coil map is imperfect, such as under exercise stress, deep breathing, and patient motion.

4.2 The Use of a 32-Channel Cardiac Coil to Improve the Temporal Resolution

We have shown in Section 2.8.2 based on the literature findings that a high temporal resolution is required for accurate quantification of left ventricular volumes, as shown in Figure 16 and Figure 17 as a function of age. Depending on the level of acceptable error, for a 50 year old individual, temporal resolution ranging from 51.9 ms to 26.0 ms may be needed for accurate volume quantification. Therefore, higher acceleration factors are required to improve the temporal resolution. A higher number of cardiac array coil elements would increase the number of equations available to reconstruct the under-sampled data, and reduce the spatially dependent noise.
amplification due to ill-conditioning in the reconstruction (g-factor). As a result, we investigated whether a 32-channel cardiac array coil can enable rate 4 parallel acceleration at peak stress. Rate 4 has previously been reported as the optimal 1D acceleration rate with the 32-channel cardiac array coil under resting conditions; at higher acceleration rates, the g-factor was found to significantly increase [95].

4.2.1 Methods

Six subjects exercised on a fully MRI-compatible hydraulic treadmill described in detail in Chapter 7, which was positioned immediately adjacent to the MRI table, analogous to a stress echocardiography lab. Cardiac cine data was obtained as quickly as possible post-exercise, including 3 short-axis, 1 vertical long-axis, and 1 horizontal long-axis slices at 2.2 seconds per slice using a 32-channel cardiac array coil. Cines were acquired using both rate 3 and rate 4 parallel acceleration with TGRAPPA reconstruction (TR/TE 2.3/1.0 ms, matrix 84x160, average FOV 306x383 mm) and all frames were averaged to obtain the coil sensitivity maps. The order of acquisition was randomized such that in 3 subjects, rate 4 cines were acquired first, and in the other 3 subjects, rate 3 cines were acquired first. The average temporal resolution at rate 4 was 47.6±1.3 ms compared to 63.5±1.7 ms at rate 3.

The mid short-axis slice, and the vertical and horizontal long-axis slices of each subject at rate 3 and rate 4 were anonymized and randomized, and were reviewed by an
experienced cardiologist blinded to the study. An image quality score was assigned to
each of the 36 cines according to the following scale: (1) excellent, (2) good, (3)
diagnosis may be limited, and (4) poor, non-diagnostic.

4.2.2 Results and Conclusions

The average image quality score at rate 3 was 2.0±0.9 and at rate 4 was 1.9±0.9.
A comparative example showing multiple views is presented in Figure 41. Therefore, rate
4 parallel acceleration is feasible at peak exercise stress using the 32-channel cardiac
array coil without impairing the diagnostic quality of the images. This enables a higher
temporal resolution to be achieved for more accurate quantification of left ventricular
volumes that are used to compute the ejection fraction (EF), which is an important tool in
determining the impact of exercise on the global cardiac function.
Figure 41: Rate 3 (top row) and rate 4 (bottom row) images acquired in the same subject following peak exercise. Short-axis (left), vertical long-axis (middle), and horizontal long-axis (right) views are shown.
CHAPTER 5

MAGNETIC FIELD THRESHOLD FOR ACCURATE
ELECTROCARDIOGRAPHY IN THE MRI ENVIRONMENT

Accurate 12-lead ECG monitoring is required both during exercise stress testing and recovery. Since exercise CMR is performed inside the MRI room, the 12-lead ECG accuracy may be adversely affected by the magnetic field. Although the ECG is known to be non-diagnostic within the bore of any high-field magnet due to the magnetohydrodynamic (MHD) effect, it may be feasible inside the MRI room but outside of the bore. The aim of this chapter is to quantify the magnetic field threshold for accurate ECG monitoring. In order to accomplish this, the MHD effects on the ST segment of the ECG waveform are investigated in 6 subjects lying supine on the MRI table with the aortic mid-arch positioned at distances from the magnet bore entrance ranging from 182 to 26 cm. The resulting threshold for accurate ECG is presented, and related to subject positioning on the MRI table and treadmill positioning adjacent to the MRI table.
5.1 Introduction

The electrocardiogram (ECG) is known to be significantly distorted by the magnetohydrodynamic (MHD) effect and is non-diagnostic within the bore of any MRI magnet. The MHD effect results from blood flow within the static magnetic field and is most pronounced when flow is rapid and oriented perpendicular to the magnetic field, e.g., in the aortic arch [56]. While techniques for R-wave detection in the MRI bore are suitable for ECG triggering of MR data acquisition [90], accurate ECG monitoring is not feasible due to MHD distortion. Knowledge of the magnetic field threshold at which distortion of the ECG becomes significant is important to ensure the accuracy of ECG monitoring within the fringe field inside the MRI room, but outside of the magnet bore. Continuous 12-lead ECG monitoring is required during and immediately after both exercise [96] and dobutamine stress testing [97]. ECG monitoring is essential for the safety of critically ill or anesthetized patients undergoing MRI procedures, and is also vital to several new and emerging MRI applications. For example, the growing fields of interventional and intra-operative MRI are expanding the needs for advanced patient monitoring in the MRI magnet room.

As discussed in Chapter 2, exercise CMR can only be performed successfully if MRI-compatible exercise equipment is positioned on or immediately adjacent to the MRI table, necessitating accurate ECG monitoring in close proximity to the MRI magnet. Exercise-stress induced changes in the ST-segment of the ECG are indicative of ischemia, with ST depression $\geq 0.10$ mV or ST elevation $>0.10$ mV considered an
abnormal response [51]. Unfortunately, peak aortic arch flow occurs on average between 92 and 107 ms after the R-wave [98], coincident with the ST-segment, and the resulting MHD effects may mask ischemia-induced changes in the ECG. Supine [53] or upright [58] bicycle exercise inside the magnet bore precludes accurate ECG monitoring, but monitoring during bicycle exercise outside the bore on the extended MRI patient table may be feasible. ECG monitoring during exercise on a treadmill placed immediately adjacent to the MRI table may also be feasible. In addition to continuous monitoring during stress testing, ECG monitoring should resume as quickly as possible after post-stress imaging, ideally while the patient is still on the MRI table.

The American Heart Association (AHA) guidelines for automated electrocardiography [99] recommend that deviation from the true waveform for accurate visual assessment of the ECG signal may not exceed 0.025 mV or 5%, whichever is greater. The objective of the work presented in this paper is to determine the magnetic field threshold below which the MHD effects are within the AHA guideline for allowable ECG distortion. With knowledge of the fringe field of any particular MRI system, the magnetic field threshold determined in this study can be translated to an acceptable distance from the magnet where accurate ECG monitoring can be performed.
5.2 Methods

All experiments were performed on a 1.5T MRI system (MAGNETOM Avanto, Siemens Medical Solutions, Erlangen, Germany). Using a gaussmeter axial probe (LakeShore Model 420), we measured the static magnetic field (B) at the level of the patient table in 5 cm increments from the end of the fully extended table (240 cm from the bore entrance) to the magnet isocenter. A gaussmeter relies on the Hall effect to measure the fringe field, using a current passing through a flat conductor located inside the magnetic field. The magnetic field exerts a transverse force on the moving charges, pushing them to one side of the conductor, and generating a potential difference which is linearly proportional to the magnetic field strength.

Assuming that aortic arch flow is the primary source of the MHD effect [56], we used MRI scout images to find the distance from the aortic arch to the isocenter in all subjects in order to determine the magnetic field at the aortic arch at various table positions. The induced MHD voltage across the aorta may be expressed as [56]:

\[ V = \int u \times B \cdot dL \]  
\[ \text{Equation 3} \]

where \( u \) is the blood velocity (m/s), \( B \) the magnetic flux density (T), and \( L \) is the aortic diameter (m). This can be simplified to:

\[ V = uBL \sin(\theta) \]  
\[ \text{Equation 4} \]
where $\theta$ is the angle between the magnetic field vector and the direction of flow. The greatest MHD voltage is induced when the magnetic field is perpendicular to the direction of flow. This relationship also indicates that the MHD signal should be linearly proportional to $B$ for any subject. Lead I is expected to have the strongest MHD effect among the 12 ECG leads due to its geometric orientation, which is approximately perpendicular to both the static magnetic field and the aortic arch. Lead I voltage is defined as the difference in potential between the left arm (LA) and right arm (RA), while the right leg (RL) electrode is the ground. In order to minimize artifact due to limb motion during exercise, the AHA guidelines recommend an alternative placement of the limb leads on the subject’s torso, as illustrated in Figure 42.

Figure 42: Electrode placement for Lead I of a 12-lead electrocardiogram
We recorded ECG data in 6 healthy subjects (ages 21 to 29) lying supine on the MRI table as well as on a table outside of the MRI room. All subjects gave written informed consent. The exclusion criteria were known or suspected cardiovascular disease and the standard contraindications to MRI. In each subject we acquired 2 minutes of supine ECG data using a 12-lead ECG system (MP100A-CE, Biopac, Santa Barbara, CA) at a 1kHz sampling rate. The Biopac system was selected to perform the data acquisition instead of the devices used for ECG gating during MRI imaging because these utilize filtering that may alter the MHD signal.

The measurements were performed at 4 to 6 table positions (depending on the subject’s height) with the subject feet-first toward the magnet, starting with the table fully extended and moving the table into the magnet bore. ECG signals were recorded at magnetic field strengths ranging from 6.4 mT to 652 mT with the aortic arch positioned from 182 cm to 26 cm from the bore entrance. The subjects were instructed to remain completely still during the 2-minute measurements. We also recorded the ECG of each subject while lying supine on a table outside of the MRI room to serve as a “baseline” signal with no magnetic interference.

In each subject we acquired standard scout images to determine the location of the aortic arch relative to magnet isocenter, and through-plane aortic velocity measurements perpendicular to the mid-arch (segmented k-space spoiled gradient echo, TE/TR 2.0/48.3 ms, 6.0 mm slice, matrix 100x192, and rate 2 parallel acceleration). The aortic diameter at the arch, and the primary direction of flow relative to the magnetic field were also
determined from the phase-velocity and scout images. The phase-velocity images were also analyzed using Argus software (Siemens, Malvern, PA) to determine the peak aortic arch velocity and its timing relative to the R-wave.

Data analysis was performed using MATLAB. We identified the peak of each R-wave and the corresponding T-wave, and segregated the data into individual RT (peak-to-peak) intervals. We rejected the RT intervals that fell outside of ±1 SD of the mean RT interval duration to ensure physiologic consistency, and linearly expanded or contracted the duration of each RT interval to the mean duration at baseline. We subsequently averaged all heartbeats to obtain the mean RT interval waveform at each table position. According to the AHA exercise testing standards [51], ST displacement should be measured at 60 ms or 80 ms (depending on heart rate) following the J-point, the transition between the QRS complex and ST segment. We visually identified the J-point for each mean waveform, and analyzed the subsequent 120 ms interval; this corresponds to the estimated upper limit of the ST segment duration [100, 101].

We subtracted the baseline mean ST segment from the mean ST segment at each table position in order to determine the magnitude of the MHD effect as a function of field strength. We identified the peak MHD deviation at each position in terms of the greatest absolute voltage difference from baseline. Finally, we evaluated whether the peak deviation exceeded the AHA guideline of 0.025 mV or 5%, and pooled the data from all subjects to determine the threshold at which the magnetic field begins to have a significant impact on the ECG signal.
Linear regression was performed to investigate the correlation between MHD voltages and magnetic field, between measured MHD voltages and peak aortic velocity, and between measured MHD and the voltages determined by Equation 4 in each subject. P-values less than an alpha of 0.05 were considered to indicate statistical significance.

5.3 Results

The Siemens 1.5T Avanto magnetic field as a function of distance from the bore entrance is displayed in Figure 43, with vertical lines indicating the magnet isocenter and the end of the MRI patient table when fully extended. These measurements are in agreement with the two-dimensional magnetic field plot provided by the vendor (Figure 44), noting that the vendor plot is referenced to the isocenter.
Figure 43: Measured magnetic field of the 1.5T Siemens Avanto as a function of distance from the bore entrance.

Figure 44: Two-dimensional magnetic field plot of the 1.5T Siemens Avanto provided by the vendor.
The mean RT intervals (peak-to-peak) in two subjects are displayed in Figure 45 (a and b), with the J-point indicated on each. This figure illustrates the increasing deviation from baseline at higher magnetic field strengths. The deviation from baseline within the 120 ms interval beginning at the J-point is shown in Figure 45 (c and d) for the same two subjects at various magnetic field strengths. The peak deviation from baseline is plotted versus magnetic field strength in Figure 45 (e and f), illustrating the linear relationship between the MHD effect and the magnetic field, and the difference in slope between these two subjects. The linear relationship between field strength and MHD, as well as the timings of the J-point and peak aortic velocity for all subjects are listed in Table 5.
Figure 45: (a) and (b) Mean RT intervals (peak-to-peak) in two subjects outside the MRI room (baseline) and at various magnetic field strengths. The J-point is shown (arrow). (c) and (d) The corresponding deviation from baseline at different field strengths within a 120 ms interval following the J-point. (e) and (f) Peak deviation from baseline as a function of field strength, indicating a linear relationship between the MHD effect and the magnetic field, and a difference in slope between the two subjects.
Table 5: Time following the R-wave when the J-point occurred, the correlation coefficient indicating a linear relationship between the MHD effect and the magnetic field strength, the slope of the curve representing the deviation from baseline as a function of the magnetic field strength, and the peak velocity in the aortic arch and its timing relative to the R-wave for the 6 subjects.

<table>
<thead>
<tr>
<th>Subject</th>
<th>J-point (ms)</th>
<th>Time of Peak Velocity in Aortic Arch (ms)</th>
<th>Peak Velocity in Aortic Arch (cm/s)</th>
<th>Correlation Coefficient</th>
<th>Slope (mV/T)</th>
</tr>
</thead>
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<tr>
<td>1</td>
<td>50</td>
<td>144</td>
<td>102</td>
<td>0.997, p=0.0026</td>
<td>0.936</td>
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<td>0.952, p=0.0034</td>
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<tr>
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<td>0.981, p&lt;0.001</td>
<td>0.252</td>
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<td>0.980, p&lt;0.001</td>
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<tr>
<td>6</td>
<td>50</td>
<td>97</td>
<td>98</td>
<td>0.992, p&lt;0.001</td>
<td>0.742</td>
</tr>
</tbody>
</table>

The correlation coefficient (r) between measured MHD voltages and peak aortic velocity was 0.165 with p=0.359, indicating lack of linear correlation. The slope of the regression line between measured MHD voltage and that calculated using Equation 4 was 0.046±0.038 (r=0.834, p<0.001). As expected, the measured MHD effect was much lower since the equation only predicts voltage across the aorta, and not at the chest.

The high correlation coefficients in Table 1 within the 120 ms interval indicate that the MHD effect is linearly proportional to the magnetic field in each subject. Figure 46a displays the peak deviation from baseline for all 6 subjects at all measured positions. The circles indicate the points within 0.025 mV or 5%, while the squares indicate those which exceed this AHA guideline for maximum allowed signal distortion. This plot
shows that all points below a magnetic field strength of 70.7 mT are within the 0.025 mV or 5% guideline. Figure 46b shows the corresponding plot of peak deviations from baseline at various positions of the aortic arch relative to the bore entrance. For the Siemens Avanto 1.5T system used in this experiment, when the aortic arch was positioned 77 cm or more from the bore entrance, signal distortion was within the 0.025 mV or 5% guideline for all subjects.

Figure 46: (a) Peak deviation from baseline for all 6 subjects expressed as a function of magnetic field strength. Below the field strength of 70 mT, peak deviation is below 0.025 mV in all subjects. (b) Peak deviation from baseline expressed as a function of the distance of the aortic arch from the bore entrance. At distances of 80 cm or more from the bore, peak deviation is below 0.025 mV in all subjects.
Discussion and Conclusions

The aim of this study was to determine the magnetic field threshold for accurate ECG monitoring inside the MRI room. We have shown that reliable ECG measurements can be obtained within the ST segment at magnetic field strengths below approximately 70 mT measured at the aortic arch in supine subjects. With knowledge of the magnetic field plot of a particular MRI room, obtained either through gaussmeter measurements or from magnet manufacturer specifications, it is possible to determine the location relative to the MRI system that defines the 70 mT threshold. For our 1.5T system this distance is approximately 80 cm.

These results indicate that efforts should be made to either position the patient as far from the bore as possible in the head-first orientation, or position the patient feet-first toward the magnet. With the table fully extended and the subjects positioned feet-first towards the magnet, the closest aortic arch distance from the bore entrance was 182 cm, well outside the 80 cm limit. Accurate ECG recording should be feasible during exercise on a treadmill positioned adjacent to the MRI table, during recovery on the MRI table following exercise or pharmacologic stress testing, during supine bicycle exercise on the fully extended MRI table, and for general patient monitoring inside the MRI room. Although the peak aortic flow and the static magnetic field would be oriented differently during upright treadmill or bicycle exercise, our measurements represent the worst-case scenario with the flow velocity, the static magnetic field, and the ECG lead all approximately perpendicular to each other. Whereas other segments of the aorta may be
oriented perpendicular to the magnetic field when standing or sitting, the flow velocities do not vary appreciably, with average velocities of 104, 101, and 113 cm/s in the ascending, descending, and abdominal aorta, respectively [98]. A patient laying supine on the MRI table in the feet-first orientation and the fully MRI-compatible treadmill which will be described in Chapter 7 are schematically illustrated in Figure 47, depicting the 70 mT field line. The treadmill is shown with the front right up to the magnet and the side immediately adjacent to the patient table. The center of the treadmill belt located at 113.5 cm from the bore entrance is illustrated with the cross-hairs. This figure clearly shows that accurate 12-lead ECG monitoring is feasible both during exercise and recovery in the MRI room.

Figure 47: Magnetic field plot of Siemens 1.5T Avanto showing that accurate 12-lead ECG monitoring is feasible during exercise inside the MRI room and recovery on the MRI patient table with the aortic arch positioned at field strengths below 70 mT
The ECG signal is subject to worsening MHD effect at fields above 1.5T [102]. Based on the manufacturer-supplied field plots, we estimate that for the Siemens 3T Trio, the 70 mT threshold is located approximately 130 cm from the bore entrance [103], and for the Siemens 3T Verio, it is located 110 cm from the bore entrance [104]. This indicates that ECG monitoring should also be feasible at 3T with the patient positioned feet-first on the fully-extended MRI table.

The AHA guidelines specify that an acceptable threshold for ECG signal distortion due to filtering is 0.025 mV or 5%, whichever is greater. Due to the low voltages within the ST segment, 0.025 mV was the higher threshold in each case. This AHA standard defines the limit of distortion introduced by filtering as the maximum allowed deviation from the true waveform. However, in this experiment we encountered the additional factor of physiologic variability between measurements at different table positions. For example, subjects may have experienced excitation or stress as they were moved closer to the bore, altering signal amplitudes and timing. Heart rates and breathing patterns were observed to change in some instances. These physiological changes may have introduced distortion of the ECG relative to the baseline signal unrelated to the MHD effect. We may be conservative in our assessment of the relationship between field strength and MHD distortion since we are accounting for both the MHD effect and physiologic variability, although the methods of signal processing and analysis we employed were designed to minimize this effect.
The peak blood flow velocity in the aortic arch was found to occur between 109 ms and 144 ms following the R-wave in the 6 subjects. Since the J-point occurred no later than 54 ms following the R-wave, the peak blood flow velocity in these subjects occurred within the ST segment. We evaluated every point within the ST segment in 1 ms increments, and determined the maximum MHD effect. The peak deviation from the baseline ECG did not occur at the same time point at the different field strengths in each subject due to physiologic variability.

The MHD effect for each subject was linearly proportional to the magnetic field strength. The MHD plots for different subjects, such as Figure 45e and f, had different slopes due to the variability in aortic velocity, aortic arch angle and diameter, torso size, and electrode placement. The measured MHD voltages were not linearly correlated with aortic velocity (r = 0.165, p=0.359). While faster blood flow is expected to increase the induced voltage, the other effects listed above may have been dominant. While the predicted and measured induced voltages were strongly linearly correlated (r = 0.834, p<0.001), Equation 4 is an oversimplification and cannot account for the complex nature of blood flow in the heart and thoracic vasculature.

The average resting supine peak aortic flow velocity in our subjects (86 cm/s) was approximately equal to the average peak aortic velocity (80 cm/s) measured in subjects in the 50 to 74 year age range immediately following maximal treadmill exercise [105]. Therefore, these data acquired in young healthy subjects should extrapolate to the typical
cardiac patient cohort under stress conditions. At reduced cardiac output and flow velocities, the MHD effects would only be lower than those we observed.
EXERCISE CMR IN PATIENTS WITH KNOWN OR SUSPECTED CAD

The objective of this study was to test the feasibility of exercise CMR for diagnosing ischemia in patients referred for treadmill nuclear exercise testing as a result of known or suspected coronary artery disease. Coronary angiography was used as the gold standard, while long-term follow-up was used to assess outcome in patients not referred for coronary angiography. The patients exercised only once on the treadmill in the corner of the MRI room, followed by stress CMR imaging and then nuclear stress imaging. The diagnosis with each modality was made based on aggregate imaging and exercise test parameters, and compared to the results of coronary angiography or long-term follow-up.

6.1 Subjects

Forty three patients age 25 to 81 years (average age 54 ± 12) who were referred for treadmill nuclear scintigraphy because of known or suspected CAD were enrolled. The study protocol was approved by the Institutional Review Board at The Ohio State
University, and all participants gave written informed consent. The exclusion criteria were inability to exercise and standard contraindications to MRI. The decision to perform maximal treadmill exercise combined with nuclear scintigraphy was made independently of CMR. The CMR study was incorporated into the nuclear stress protocol without added risk to patients. Patients were subsequently referred for invasive coronary angiography based on physician discretion and interpretation of nuclear imaging results, ECG findings, exercise parameters, and clinical history. In the subset of patients referred for coronary angiography, the angiography findings were used as the “gold standard”, with the presence of >70% stenosis in any coronary territory defined as a positive result. Six-month follow-up was used to determine outcome in patients not referred for coronary angiography.

6.2 Combined Exercise CMR and Nuclear Protocol

The combined exercise CMR and nuclear protocol is diagramed in Figure 48. The CMR component of the test was described in more detail in Section 3.5. The patients began with routine resting nuclear scintigraphy, which lasted approximately 45 – 60 minutes. After the resting nuclear images were reviewed and the patients were cleared to proceed by the nuclear staff, they were escorted down the hall to the MRI exam room. At peak exercise, the nuclear radioisotope was injected by a nuclear medicine technologist, and the patients were immediately placed back into the MRI system and imaged. This entire procedure lasted less than 30 minutes following exercise, at which point the patients were escorted back to the nuclear camera for imaging. The pharmacokinetics of
the radiopharmaceutical do not require that stress nuclear imaging be completed immediately following exercise. In standard clinical nuclear stress testing protocols, the patients recover for 15-30 minutes before stress nuclear images are acquired. Therefore, the CMR protocol was designed to not cause additional delay between exercise and nuclear imaging. The advantage of the combined protocol is that the patients were not required to undergo two separate exercise tests (nuclear and CMR), but exercised only once inside the MRI room.

Figure 48: Combined exercise CMR / nuclear scintigraphy protocol

An adequate stress test was defined as reaching 90% of MPHR (0.9*(220 - age)). The American Heart Association standards for exercise testing [51] specify the absolute and relative indications for terminating an exercise test prior to reaching the target heart rate. The absolute indications are ST segment elevation > 1mm on the ECG, persistent drop in systolic blood pressure >10 mmHg despite the increase in workload when accompanied by any other evidence of ischemia, moderate to severe angina, central nervous system symptoms, signs of poor perfusion, sustained ventricular tachycardia, technical difficulties monitoring the ECG or blood pressure, and the subject’s request to
stop. The relative indications for terminating an exercise test are ST or QRS changes such as excessive ST displacement on the ECG, persistent drop in systolic blood pressure >10 mmHg despite an increase in workload in the absence of other evidence of ischemia, increasing chest pain, fatigue, shortness of breath, wheezing, leg cramps, or claudication, arrhythmias other than sustained ventricular tachycardia, hypertensive response (systolic blood pressure >250 mmHg and/or diastolic blood pressure >115 mmHg), development of bundle-branch block that cannot be distinguished from ventricular tachycardia, and the subject’s general appearance.

In addition, the time between the end of exercise and the start of CMR imaging, as well as the time to complete cardiac cine imaging, were recorded.

6.3 Coronary Angiography and Patient Follow-Up

The results of coronary angiography were recorded as maximum percent stenosis in the left mainstem, left anterior descending (LAD), right (RCA), and left circumflex (LCx) coronary arteries. A positive result was defined as >70% stenosis in any coronary territory.

In the patients not referred for coronary angiography, six-month follow-up from the time of the exercise CMR exam was performed by telephone contact with the patient or family member, communication with primary care physician, and/or through medical record review. Cardiovascular event was defined as death due to cardiovascular disease,
new myocardial infarction, revascularization, or a subsequent coronary angiography exam indicating >70% coronary stenosis.

In addition, there was a high pre-test likelihood that some angiography results would be negative. Attenuation artifact and other limitations of nuclear stress imaging result in occasional false positive findings in patients with no obstructive coronary disease, ranging from 2% to 19% [14], who subsequently undergo angiography based on these findings. This factor was expected to lead to corroboration of negative exercise CMR tests with angiography in some subjects.

In yet other individuals, stress testing identifies microvascular disease that is not associated with epicardial coronary artery obstruction. This difference between physiologic assessment for ischemia with stress testing and anatomic assessment for CAD with angiography could potentially occur in the study population.

6.4 Data Analysis

Exercise nuclear and CMR examinations including aggregate assessment of exercise parameters, ECG findings, myocardial perfusion, left ventricular wall motion (CMR only), and viability were independently reviewed offline by a consensus of two reviewers blinded to the results of the other imaging study, and each test was classified as either negative/adequate stress, negative/inadequate stress, positive for ischemia, or fixed abnormality/no ischemia. Patients were referred for coronary angiography based on the
nuclear images, ECG findings, exercise parameters, and clinical history, or based on physician discretion.

6.5 Results

Treadmill exercise stress was terminated due to achieving 90% of MPHR in 17 patients (40%), dyspnea in 11 patients (25%), fatigue in 11 patients (25%), chest pain in 2 patients (5%), and musculoskeletal pain in 2 patients (5%). Of these, 8 patients underwent invasive coronary angiography. Mean time to completion of cine MRI post-exercise was 68±14 sec, and to completion of perfusion imaging 88±8 sec. Accuracy in the 8 patients who underwent coronary angiography was 7/8 for CMR (1 false positive) and 5/8 for SPECT (2 false positive and 1 false negative). The 2 SPECT false positives had angiographically normal coronary arteries, and were correctly diagnosed by CMR as negative. The patient who was false negative with SPECT had >70% stenosis requiring revascularization, and was correctly diagnosed by CMR as positive for CAD. The false positive by CMR had no epicardial stenosis, but had diffuse subendocardial ischemia thought to represent microvascular disease. However, the study criteria only looked at >70% coronary stenosis as positive corroboration by angiography. Follow-up at a median of 6 months indicated freedom from cardiovascular events in 29/29 CMR-negative and 33/34 SPECT-negative patients. The difference in the number of CMR-negative and SPECT-negative patients is due to the fact that CMR identified non-transmural infarct scar in 4 patients with normal stress SPECT exams.
In the example shown in Figure 49, the abnormal CMR wall motion and perfusion findings were corroborated by coronary angiography. The nuclear perfusion images were obscured by adjacent gut uptake of the radioisotope, resulting in ambiguous findings.

Figure 49: Invasive coronary angiogram confirming multiple coronary artery blockages (arrows) responsible for findings on stress imaging. CMR shows ischemia as well as prior infarction. Nuclear images obscured by adjacent gut uptake (open arrow)

The example in Figure 50 shows a patient whose occluded right coronary artery resulted in a perfusion defect and a corresponding wall motion abnormality in exercise CMR. Nuclear images, on the other hand, were interpreted as negative; the nuclear stress perfusion defect was interpreted as attenuation artifact.
Figure 50: Invasive coronary angiogram showing occluded right coronary artery responsible for findings on stress CMR. CMR shows ischemia as inferior perfusion and wall motion defect. Nuclear defect interpreted as attenuation artifact (open arrow)

The accuracy of exercise CMR and SPECT exams are summarized in Table 6.

Table 6: Summary of accuracy findings for exercise CMR and SPECT exams. Both exams were performed in each subject at the same physiologic condition.

<table>
<thead>
<tr>
<th></th>
<th>CMR</th>
<th>Nuclear</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. of True Positives (TP)</td>
<td>7</td>
<td>5</td>
</tr>
<tr>
<td>No. of False Negatives (FN)</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>No. of True Negatives (TN)</td>
<td>29</td>
<td>33</td>
</tr>
<tr>
<td>No. of False Positives (FP)</td>
<td>1</td>
<td>2</td>
</tr>
</tbody>
</table>

6.6 Discussion

We have demonstrated that exercise CMR is feasible for accurately detecting CAD, with fewer false positives and false negatives than stress SPECT. Although these
results appear highly promising, we do not currently have the required sample size to perform a direct statistical comparison between the two modalities in order to draw statistical conclusions.

Based on data from previous published studies, we expect the sensitivity and specificity of SPECT to be approximately 88% and 77%, respectively [10], while the sensitivity and specificity of CMR should be at least 89% and 87% [12] based on pharmacological stress CMR of myocardial perfusion and viability. In order to provide 80% power to detect a difference in specificity of 10% between SPECT and CMR at a two-sided alpha of 0.05, we will need to enroll 181 CAD negative patients. Assuming a patient population that is 80% CAD negative, this will require a total enrollment of 227 patients. Therefore, we will need to enroll 184 more patients.

In addition to providing a viable (and potentially superior) alternative to nuclear stress imaging, exercise CMR may also address the limitations of stress echocardiography outlined in Chapter 2. Nuclear stress perfusion imaging was chosen for this comparison because the cardiomyocyte intracellular uptake and retention of the radioisotope “freezes” the stress perfusion state and allows an extended period (15-30 min) between the isotope injection at peak stress and nuclear stress imaging. This enables stress CMR function, perfusion, and viability imaging to be performed in this time interval. The CMR study in no way interferes with the clinical nuclear exam; the only modification is that the treadmill is located inside the MRI room. Similarly, nuclear imaging does not alter the CMR protocol except for radioisotope injection during
exercise. In addition to reducing patient risk by exercising them only once, this also ensures that the subjects are at the same physiologic stress condition with both modalities. A direct comparison of exercise CMR with any other modality would require two stress tests, and therefore increase patient risk while potentially introducing physiologic variability.

The primary challenge in obtaining specificity values was the fact that patients with negative stress tests typically do not undergo invasive coronary angiography (the gold standard). However, in this subset of patients, six-month follow-up was used to assess outcome. In order to further improve the assessment of exercise CMR accuracy in ruling out CAD, the follow-up will be repeated one year after the exercise CMR exam. Furthermore, pharmacologic stress CMR studies show that an abnormal CMR adds significant prognostic value for CAD, myocardial infraction, or death over clinical risk factors [106]. While coronary angiography shows the vessel narrowing, this is not always related to functional significance of disease. Therefore, one-year follow-up will be performed in all patients in order to investigate the prognostic value of exercise CMR. Similar one to two-year follow-up studies for CMR have been reported in literature [106, 107].
6.7 Conclusions

Exercise stress CMR including wall motion and perfusion inside the MRI room is feasible in patients with known or suspected ischemic heart disease. Preliminary results indicate favorable accuracy of this new stress imaging system compared to nuclear perfusion imaging. However, the sample size will need to be increased in order to draw statistical conclusions, which will require a larger clinical trial. Furthermore, in order to expedite the completion of cine and perfusion imaging as quickly as possible after peak exercise and ensure patient safety, the treadmill in the corner of the MRI room will need to be replaced by a fully MRI-compatible treadmill positioned immediately adjacent to the MRI table.
CHAPTER 7

DEVELOPMENT AND VALIDATION OF THE CONTROL SYSTEM FOR A FULLY MRI-COMPATIBLE HYDRAULIC TREADMILL

This chapter will focus on the development and testing of the control system for an MRI-compatible hydraulic treadmill which can operate immediately adjacent to the MRI table. As discussed in Chapter 2, the objective is to enable imaging to commence rapidly after the completion of exercise in order to image the heart as close to peak stress as possible.

Section 7.1 will provide a broad overview of the MRI-compatible treadmill system, while the remainder of this chapter will specifically focus on all aspects pertaining to the development and testing of the feedback control system for continuous control of speed and elevation. Component selection and the design of speed and elevation control will be described. The LabVIEW software design and implementation, and the failure modes and effects analysis and the implementation of safety features will be discussed next. The results of control system performance testing across the range of speeds and elevations corresponding to six stages of the Bruce treadmill protocol will be
presented. Finally, testing in subjects undergoing the Bruce protocol to reach peak stress immediately adjacent to the MRI table will be presented.

7.1 MRI-Compatible Treadmill System Overview

The MRI-compatible treadmill system is illustrated in the schematic in Figure 51. An electric motor located outside of the MRI room drives a hydraulic pump (Figure 52), which sends water using 50-ft hoses through a waveguide to an MRI-compatible hydraulic motor located on the treadmill inside the MRI room. The resulting flow across the hydraulic motor causes the drive shaft to turn. The drive shaft is connected to a flywheel, which dampens sudden changes in loading and acceleration, and is also connected to the front roller of the treadmill via a drive belt (Figure 53). The front roller causes the treadmill belt to turn. A hydraulic accumulator located outside of the MRI room supplies water to a non-ferromagnetic lift cylinder on the treadmill through a 50-ft hose. Raising the treadmill is accomplished by discharging water from the accumulator into the lift cylinder, while lowering the treadmill occurs by allowing the fluid from the cylinder to drain back into the reservoir using the force of gravity. More details on the mechanical and hydraulic systems may be found in a Master’s Thesis by Foster, E. [108].
Figure 51: Schematic of the MRI-compatible treadmill system

Figure 52: Electrical and hydraulic power components located outside the MRI room, which are connected to the treadmill via hydraulic hoses
The elevation system contains three solenoid valves. The “accumulator” valve function is to charge the accumulator, i.e. fill it with water which can subsequently be released into the lift cylinder to raise the treadmill during a test. The accumulator valve controls the direction of flow from the hydraulic pump into either the hydraulic motor or the accumulator, as illustrated in Figure 54. While the accumulator is being charged, there is no water flow to the hydraulic motor and the treadmill belt is thus stationary.
Figure 54: Accumulator valve diverts flow from the hydraulic pump into either the hydraulic motor or the accumulator

The operation of the other two solenoids (the “raise” valve and the “lower” valve) is illustrated in Figure 55, which shows both solenoids in “closed” configuration. In “open” configuration, there would be no path for water to flow, such that the cylinder stroke would remain constant and the treadmill would maintain its current elevation. This state requires power to be supplied to both solenoids. When no power is supplied, the solenoids are in “closed” configuration and allow the water to flow from both the cylinder and the accumulator back into the reservoir, which is the path of least resistance. In this case, the treadmill lowers fully. When the raise valve is closed and the lower valve is open, the water flows from the pressurized accumulator into one side of the lift cylinder. This causes the cylinder rod to move to the right (based on the orientation in the schematic) and shorten the cylinder stroke, and moves the treadmill leg assembly coupled to the cylinder to raise the treadmill. On the other hand, when the raise valve is open and the lower valve is closed, the water flows from the lift cylinder back into the reservoir, causing the cylinder rod to move to the left, lengthening the cylinder stroke, and moving the leg assembly to lower the treadmill. The accumulator charge remains constant in this
case since the raise valve prevents discharge into the reservoir. The state of all three solenoid valves can be changed digitally using commands sent from the LabVIEW software through a data acquisition board (DAQ). To simplify discussion in subsequent sections, especially pertaining to software control, the valve “closed” state diagrammed in Figure 55 will be referred to as the “ON” state, which requires a digital signal to be sent through the DAQ. The “OFF” state is the normal state when power is on and no commands are sent to the valves. The effect of these states on the treadmill and the accumulator is summarized in Table 7.

Figure 55: The operation of the “raise” and “lower” valves, which control the treadmill elevation by adjusting the cylinder stroke and moving the leg assembly
Table 7: The combined effect of the raise and lower valves on the treadmill and the accumulator

<table>
<thead>
<tr>
<th>Raise valve</th>
<th>Lower valve</th>
<th>Effect on treadmill</th>
<th>Effect on accumulator</th>
</tr>
</thead>
<tbody>
<tr>
<td>ON</td>
<td>ON</td>
<td>Lowers fully</td>
<td>Discharges fully</td>
</tr>
<tr>
<td>OFF</td>
<td>ON</td>
<td>Lowers</td>
<td>Maintains charge</td>
</tr>
<tr>
<td>ON</td>
<td>OFF</td>
<td>Raises</td>
<td>Decreases</td>
</tr>
<tr>
<td>OFF</td>
<td>OFF</td>
<td>Maintains elevation</td>
<td>Maintains charge</td>
</tr>
</tbody>
</table>

7.2 Control System Layout

The overall control system layout is displayed in Figure 56. The communication between the LabVIEW software on the PC and the various components of the control system is achieved through a National Instruments USB-6008 data acquisition board (DAQ), which is powered by the PC using the USB port. The motor controller indicated in the schematic powers and controls the electric motor connected to the hydraulic pump. These components will be described in more detail in subsequent sections. The commanded speed is sent from the PC to the motor controller, while the speed feedback is simultaneously sent to both the motor controller and the PC. The speed control is performed using the proportional-integral (PI) controller of the motor controller. The start and stop command, as well as the command to enable or disable the PI control, are sent from the PC to the motor controller. The elevation feedback from an analog output sensor is sent to the PC, and the elevation control is performed by turning on and off the raise and lower valves using the programmed control algorithm. The accumulator is charged by controlling the state of the accumulator valve. The additional elements include water
level, temperature, and accumulator pressure monitoring, an emergency stop button on
the treadmill, and motor controller wiring designed to automatically shut off the electric
motor if its speed exceeds the maximum expected value.

7.3 Speed Feedback Control System

The goal of the speed feedback control system is to enable continuous control of
the treadmill belt speed in real-time across a range of speeds up to stage 6 of the Bruce
protocol (5.5 mph) while minimizing the overshoot and steady state error. The design
goal was for the steady state error to be within 2% of the target at any speed across the
entire range. The treadmill speed and elevation are used to calculate the “metabolic
equivalents” (METS) as an estimate of the work done by a patient during treadmill exercise as follows [109]:

\[
METS = \frac{(Speed \times 26.8 \times (0.1 + \% Grade \times 0.018) + 3.5)}{3.5}
\]  

Equation 5

The basal metabolic rate is defined as 3.5 mLO2/kg/min. Table 8 displays the METS corresponding to 6 stages of the Bruce protocol, the stage-to-stage METS increase, and the change in METS resulting from a 2% speed error. A worst-case speed error of 2% results in a METS change of approximately 2%, which is relatively small compared to the stage-to-stage METS increase.

Table 8: The work done by a patient at each stage of the Bruce protocol, expressed as “metabolic equivalents” (METS). The stage-to-stage increase in METS, and the error in METS resulting from a speed error of 2% are also shown.

<table>
<thead>
<tr>
<th>Stage</th>
</tr>
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<tbody>
<tr>
<td>1</td>
</tr>
<tr>
<td>2</td>
</tr>
<tr>
<td>3</td>
</tr>
<tr>
<td>4</td>
</tr>
<tr>
<td>5</td>
</tr>
<tr>
<td>6</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Speed (mph)</th>
<th>METS (mLO2/kg/min)</th>
<th>Stage-to-stage METS increase (%)</th>
<th>METS with 2% speed error (%)</th>
<th>METS change (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.7</td>
<td>4.64</td>
<td>N/A</td>
<td>4.72</td>
<td>1.57</td>
</tr>
<tr>
<td>2.5</td>
<td>7.05</td>
<td>51.8</td>
<td>7.17</td>
<td>1.72</td>
</tr>
<tr>
<td>3.4</td>
<td>10.16</td>
<td>44.2</td>
<td>10.35</td>
<td>1.80</td>
</tr>
<tr>
<td>4.2</td>
<td>13.48</td>
<td>32.6</td>
<td>13.73</td>
<td>1.85</td>
</tr>
<tr>
<td>5.0</td>
<td>17.23</td>
<td>27.9</td>
<td>17.56</td>
<td>1.88</td>
</tr>
<tr>
<td>5.5</td>
<td>20.37</td>
<td>18.2</td>
<td>20.76</td>
<td>1.90</td>
</tr>
</tbody>
</table>

Additional design goals include having completely non-magnetic sensing components on the treadmill, no electrical cables going in or out of the MRI room which
can introduce radiofrequency (RF) noise, and the ability to programmatically adjust the commanded speed to any desired value within the design range.

7.3.1 Selection of the Control Variable

As previously discussed, the treadmill speed system is comprised of the following sequence of components:

1. Motor controller
2. Electric motor
3. Hydraulic pump
4. Hydraulic motor
5. Flywheel
6. Drive belt and pulley
7. Front roller
8. Treadmill belt

Although the ultimate goal of the system is to control the treadmill belt speed, the flywheel frequency is easier to measure and may be used as the control variable. There is a direct relationship between the flywheel frequency and the belt speed, except for potential slippage of either the drive belt or the treadmill belt which will be evaluated during speed performance testing in Section 7.8.1. This means that the desired flywheel frequency can be determined from the desired belt speed. Once the desired flywheel frequency is compared to the feedback frequency, the electric motor can speed up or slow
down to compensate for the error. It should be noted that the flywheel speed is different from the electric motor speed since the flywheel is not directly coupled to the electric motor.

In order to determine the desired flywheel frequency from the desired belt speed, the first step was to derive the relationship between the belt speed and the front roller speed. The instantaneous belt velocity (V), the belt thickness (T_{belt}=0.1’’), and the front roller radius (R_{roller}=1.25’’) are illustrated in Figure 57.

![Figure 57: Parameters needed to compute the relationship between the belt speed and the front roller rotational frequency](image)

The equations relating the rotational frequency of the front roller to the belt speed are shown below:

\[ V = 2\pi \cdot f_{\text{roller}} \cdot (R_{\text{roller}} + \frac{T_{\text{belt}}}{2}) \]  

Equation 6
\[ f_{\text{roller}}(Hz) = \frac{V(\text{mph}) \times 1.46666(\text{ft/s})}{(R_{\text{roller}} + \frac{T_{\text{belt}}}{2}) \text{ft}} \times \frac{1}{2\pi} \]

Equation 7

The front roller is attached to a pulley, which is coupled using a drive belt to a second pulley on the drive shaft that rotates at the same frequency as the flywheel (Figure 58). Given a drive ratio of 3 between the front roller pulley and the drive shaft pulley, and substituting numbers into the above equation yields the following relationship between the flywheel frequency and belt speed:

\[ f_{\text{flywheel}}(Hz) = 3 f_{\text{roller}} = 6.464139 \times V(\text{mph}) \]

Equation 8

Figure 58: Front roller pulley, drive shaft pulley, and drive belt dimensions

The results including several intermediate parameters are shown in Table 9 for 7 stages of the Bruce protocol. These results show that the flywheel frequency ranges from 11 Hz at 1.7 mph to 38.8 Hz at 6 mph.
Table 9: Relationship between flywheel speed and belt speed at 7 stages of the Bruce protocol

<table>
<thead>
<tr>
<th>Bruce Stage</th>
<th>Belt Speed (mph)</th>
<th>Belt Speed (RPM)</th>
<th>( f_{\text{roller}} ) (Hz)</th>
<th>( f_{\text{flywheel}} ) (Hz)</th>
<th>Flywheel Speed (RPM)</th>
<th>Flywheel Speed (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.7</td>
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<td>81.23</td>
<td>243.69</td>
<td>2327</td>
<td>38.8</td>
</tr>
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</table>

7.3.2 Sensor Selection and Target Design

In order to meet the design goals, a photoelectric fiber optic switch (MiniSight, Rockwell Automation, Inc., MD) was selected to provide the speed feedback (Figure 59). The photoelectric switch is sensitive to visible light at the wavelength of 660 nm, and produces an output of 24 V if a light intensity threshold is exceeded and 0 V otherwise, thus enabling pulse detection. This switch is connected to 50-ft long plastic fiber optic cables with 1-mm core and 2.2-mm jacket (IF 121M-15-0, Industrial Fiber Optics, Inc.). The ends of the fiber cables are placed in non-magnetic sensing tips made of series 303 stainless steel (FA510, McMaster-Carr, Inc.). The fiber cables are positioned to detect the speed of flywheel rotation, as described below.
The original speed target design used strips of reflective tape placed on the flywheel, which was painted black, in order to detect transitions between light and dark bands using a duplex fiber optic cable placed in reflective mode. These pulses could then be converted to flywheel speed and thus belt speed. However, this approach had several drawbacks. Since the optical beam is an expanding cone of light, most of the light hitting the reflective surface was scattered away and only a small amount was be reflected back to the receiving fiber, thus resulting in a limited difference in light intensity between the bright and dark bands. This led to missed pulses, resulting in a low speed feedback which could cause the control system to overcompensate and the belt speed to rapidly increase. In addition, this approach was highly sensitive to the distance between the duplex fiber
and the flywheel. If positioned too close, no “OFF” states (dark bands) would be detected and if positioned too far, no “ON” states (bright bands) would be picked up by the sensor.

As a result of these drawbacks, the transmitting and receiving fibers were placed in a through-beam configuration (Figure 60). This configuration is also schematically illustrated in Figure 61. Since the light beam travels directly from the transmitter to the receiver without first having to reflect off a surface, the received light intensity is greater than in the reflective mode, producing a greater difference in light intensity between the “ON” and “OFF” states, and resulting in improved reliability. Given the sensor response time of 1 ms and the maximum flywheel speed of 38.8 Hz (25.77 ms / cycle), a target with 12 pulses per revolution (PPR) should be feasible. However, this could not be achieved experimentally because the light beam is not a “pencil” beam, but an expanding cone of light, and the beam diameter thus “blurred” the waveform. As a result, a target with 5 PPR was used instead, which provided adequate resolution even at the lowest speed of 11 Hz, as will be shown later.
7.3.3 Speed Control System Design

The speed control was performed using the Allen Bradley PowerFlex 70 motor controller (20AD8P0A3AYNAEG1, Rockwell Automation, Inc.) rated at 480 V, 8 A, and 5 horsepower (HP), which enabled proportional-integral (PI) control. The electric motor
driven by the motor controller was a 4-pole, 3-phase AC induction motor with the nameplate (NP) information listed in Appendix A (parameters 41 to 45). When an applied load slowed down the rotor of the induction motor, this created a difference (slip) between the rotor speed and the stator magnetic field rotation speed, which resulted in torque to match the load.

An optional motor controller encoder interface board, which accepts input of 5V or 12V, was selected to allow an external sensor to be used as the speed feedback signal. The 5V operation, which recognizes an input of >3.5V as “high” and <1V as “low”, was chosen instead of 12V such that the feedback signal could be simultaneously sent into the LabVIEW program through the built-in DAQ counter (PFI0). As a result, a voltage divider was constructed to reduce the sensor output from 24V down to 5V (Figure 62). Based on the schematic, the voltage input for the encoder interface board and the DAQ card was measured to be 4.6V when the sensor received light intensity exceeded the threshold. The process by which the output of the DAQ counter, which counts falling edges, was converted to flywheel frequency using the LabVIEW program will be described in Section 7.7.4.2.

![Figure 62: Voltage divider to reduce sensor output from 24 V to 5 V;](image)

\[ R1 = 2.2 \text{ k}\Omega, \quad R2 = 1 \text{ k}\Omega \]
Figure 63 shows the overall layout of the control system of the motor controller. The reference (commanded) speed is generated using the DAQ analog voltage output (AO0), and sent into the motor controller analog input 1. This voltage is proportional to the commanded frequency; the relevant motor controller parameters (126, 322, 55, 460, 461) are shown in Appendix A. Initially, the speed regulator loop was used for feedback control (Figure 63). However, this regulator is typically used when the load is directly coupled to the electric motor shaft, which is not the case in this system since the electric motor is coupled to a pump which drives water into a hydraulic motor. With this configuration, the output of the control system was inadequate to drive the load, and the belt speed was consistently below desired with a person walking or running on the treadmill. As a result, the feedback control through the speed regulator was disabled by setting it to “open loop” configuration (parameter 80 in Appendix A). Instead, the “Process PI” loop was used (Figure 63). Process PI is appropriate when an encoder is used to control an external process or speed of an external device such as the treadmill flywheel, which is not directly coupled to the electric motor shaft. The motor control parameters used in this loop are listed in Appendix A (parameters 124-133, 460-463).
A general description of proportional-integral-derivative (PID) control will be provided next, and related to the hydraulic treadmill speed control system. The general PID control schematic is illustrated in Figure 64. For the treadmill system, the control variable is the flywheel speed, and the error term results from the difference in the measured flywheel speed obtained using the speed sensor and the desired flywheel speed commanded from LabVIEW. The downstream effect of regulating the flywheel speed is the control of belt speed since the two speeds are directly proportional (except for potential belt slippage). The “output” in this case consists of three elements. The first is a change in the electric motor speed commanded by the motor controller, which elicits a change in the hydraulic pump speed. The hydraulic pump speed change in turn alters the flow across the hydraulic motor, causing the flywheel speed to change. The electric
motor speeds up or slows down until the measured flywheel speed matches the desired flywheel speed, and as such does not require knowledge of the speed of the hydraulic pump. The hydraulic motor and the hydraulic pump have a variable ratio depending on load. The output of the treadmill speed control system is the sum of the P and I terms, which will be described next. The D term will also be described, but was not applied to this system.

Figure 64: PID control schematic

Proportional Term (P)

The proportional term produces an output proportional to the error magnitude $e(t)$ and a gain $K_p$. The proportional term is expressed as:

$$P = K_p \cdot e(t)$$

where $K_p$ is the tunable proportional gain. A larger gain results in a greater change in output corresponding to a given error. If the proportional gain is too high, the system can become unstable, and if it is too low, it may not adequately respond to system
disturbances. By itself, pure proportional control never settles at the target value, and as a result is often combined with integral control. For controlling the hydraulic treadmill, the goal was to start with the default gain settings to determine if they are suitable for this application, and perform gain optimization if necessary. The default proportional gain of 1 provided by the PowerFlex 70 motor controller was found to be appropriate for this application based on the results of speed performance testing that will be described later in this chapter.

Integral Term (I)

The integral term adjusts the output based on both the duration of error (the longer an error is present, the harder it tries to correct) and the error magnitude. It is expressed as:

\[ I = Ki \int_0^{Ti} e(\tau) d\tau \]

In the PowerFlex 70, the integral term is adjusted through the integral time (Ti) over which the error is accumulated. The integral term eliminates the steady state error which occurs with pure proportional control. However, since it responds to accumulated error, including past error which is no longer present, it can cause the system to overshoot. For the hydraulic treadmill, the default integral time of 2 seconds provided by the PowerFlex 70 was found to be appropriate for this application based on the results of speed performance testing described later in this chapter.
Derivative Term (D)

The derivative term is proportional to the rate of change of error and a derivative gain $K_d$, as follows:

$$ D = K_d \frac{de(t)}{dt} $$

The derivative term may potentially reduce the amount of overshoot caused by the integral term, but is highly sensitive to measurement noise since differentiation of a signal amplifies noise. As a result, a gain too high may cause the system to become unstable. The derivative term is not available with PowerFlex 70.

The treadmill speed performance over the range of speeds corresponding to 6 stages of the Bruce protocol will be evaluated in Section 7.8.1 across several subject weights.

7.3.4 Special Case: Startup

An important issue which was encountered with the control system was the presence of significant overshoot and undershoot at belt startup from rest, which increased the risk of a person falling. This was even more pronounced due to the speed differential resulting from the fact that the undershoot immediately followed the overshoot. The hydraulic motor breakaway performance is dependent on static friction and the output of the pump and valves where leakage may occur. This breakaway
performance leads to some overshoot and subsequent undershoot, which were exacerbated by the PI control system. The proportional term attempted to compensate for the large error magnitude between the target speed and zero feedback, while the effect of the integral term was magnified by the time required to build enough pressure across the motor to start the belt, during which zero feedback was detected (for example, approximately 1 second with a 177 lb person). The startup with the treadmill at its lowest position is shown in Figure 65 with a 177 lb person, and the corresponding startup with the treadmill at stage 1 elevation is shown in Figure 66. These figures illustrate the significant overshoot (46% when treadmill is flat and 38% when it is at stage 1) and undershoot (45% when the treadmill is flat and 41% when it is at stage 1) produced by the control system. As a result, an alternate startup routine was incorporated by temporarily switching off PI control and directly commanding the desired speed to the electric motor, as will be described next.
Figure 65: Startup with a 177 lb person when the PI control system is on and the treadmill is at its lowest elevation, resulting in overshoot of 45% and undershoot of 49%.

Figure 66: Startup with a 177 lb person when the PI control system is on and the treadmill is at stage 1 elevation, resulting in overshoot of 38% and undershoot of 41%.
In order to determine the relationship between the electric motor speed and the flywheel speed, the system was taken through a range of electric motor speeds with the PI control disabled, and the corresponding flywheel speeds obtained using the speed sensor were recorded. A plot of the flywheel speed vs. the electric motor speed is presented in Figure 67, which shows a linear relationship with slope = 0.600 and intercept = -1.853. Based on this relationship, the electric motor speed was set to correspond to stage 1 by default in order to minimize adjustment once the PI control is turned back on.

![Figure 67: Flywheel speed in relation to electric motor speed with PI control disabled](image)

Figure 68 illustrates the startup without PI control when the treadmill is at its lowest elevation (overshoot of 16% and undershoot at 27%) and Figure 69 shows the startup when the treadmill is elevated to stage 1 of the Bruce protocol (overshoot of 31% and undershoot of 5%). These results show a significant improvement over startup using the control system. When the treadmill was flat, the net difference between the peak
overshoot and peak undershoot decreased from 94% of target to 43%, and when the treadmill was at stage 1, the net difference decreased from 79% of target to 36%. The time for the belt to start moving was approximately 2 seconds when the treadmill was flat and 1 second when the treadmill was elevated to stage 1.

![Figure 68: Startup with a 177 lb person when the PI control system is off and the treadmill is at its lowest elevation, resulting in overshoot of 16% and undershoot of 27%](image)

Figure 68: Startup with a 177 lb person when the PI control system is off and the treadmill is at its lowest elevation, resulting in overshoot of 16% and undershoot of 27%
Figure 69: Startup with a 177 lb person when the PI control system is off and the treadmill is at stage 1 elevation, resulting in overshoot of 31% and undershoot of 5%.

One aspect to consider with this approach is whether the belt would be able to start from rest without PI control. At stage 1 of the Bruce protocol, the presence of the 5.7° incline assists with the starting motion due to the component of weight parallel to the belt ($F_\parallel = W \times \sin(5.7°) = \sim 0.1 \times W$), as illustrated in Figure 70. At the same time, the component of weight perpendicular to the belt (the load $F_\perp$) is reduced. This is supported by the above results which show that the time to start the motor was reduced from 2 seconds when the treadmill was flat with a 177 lb subject down to 1 second. As will be shown in Section 7.9, the belt successfully started from rest with all subjects undergoing the Bruce protocol ranging in weight from 118 lbs up to 218 lbs. Although this has not happened in practice, if a situation were to occur when the belt did not start unassisted,
the worst case scenario would be that a staff member could “kick-start” the belt with their foot, thus effectively increasing $F_{ll}$.

![Figure 70: Schematic illustrating the reduced force required to move the belt when the treadmill is elevated, such as at stage 1 of the Bruce protocol](image)

Once the electric motor is started at the commanded speed without PI control, a pulse counter is used in LabVIEW to detect the point at which the flywheel starts to move. After at least 2 pulses on the flywheel are detected, the belt continues to run for another 3 seconds without PI control in order to avoid turning on PI control during the zone where potential overshoot may occur due to the hydraulic motor itself since this may cause over-compensation and exacerbate the error. For the example of Figure 68 and Figure 69, the PI control would be started at approximately 5 seconds and 4 seconds, respectively. The event flowchart is illustrated schematically in Figure 71; more information on the LabVIEW implementation is provided in Section 7.7.
7.4 Elevation Feedback Control System

The goal of the elevation feedback control system is to enable continuous control of the treadmill elevation in real-time across a range of elevations up to stage 6 of the Bruce protocol (11.3°) while minimizing the overshoot, steady state error, and time to reach the steady state. The design goal was to keep the steady state error within 3% of the target at any elevation across the entire range. The change in METS due to 3% elevation error is displayed in Table 10, which shows that the resulting METS change is approximately 2%.

Table 10: The work done by a patient at each stage of the Bruce protocol, expressed as “metabolic equivalents” (METS). The change in METS due to 3% elevation error is also shown.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Elevation (°)</th>
<th>Grade (%)</th>
<th>METS (mLO₂/kg/min)</th>
<th>METS with 3% elevation error (mLO₂/kg/min)</th>
<th>METS change (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>10</td>
<td>4.64</td>
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<td>10.2</td>
<td>18</td>
<td>17.23</td>
<td>17.61</td>
<td>2.18</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>20</td>
<td>20.37</td>
<td>20.84</td>
<td>2.29</td>
</tr>
</tbody>
</table>
Additional design goals include having completely non-magnetic sensing components on the treadmill, no electrical cables going in or out of the MRI room which can bring in RF noise, and being able to programatically adjust the commanded elevation to any desired value within the design range.

7.4.1 Sensor Selection and Target Design

In order to meet the design goals, an analog output sensor was selected which generates a voltage output proportional to the received light intensity. Similar to the speed switch, the elevation sensor is connected to 50-ft long plastic fiber optic cables (1 mm core, 2.2 mm jacket) with transmitting and receiving fibers mounted on the treadmill using non-magnetic sensing tips, and operates on a wavelength of 660 nm in the visible red range. The elevation sensing is based on the fact that there is a direct relationship between the treadmill elevation and the stroke of the lift cylinder. The primary potential sources of discrepancy include compliance introduced by air that enters the system, the compliance of hydraulic hoses which have an elastomer inner wall, and mechanical deflection such as bending in the aluminum leg assembly. The original configuration used a duplex fiber optic cable on one end of the cylinder and a reflective target on the other end, such that a changing stroke changed the distance between the fiber sensing tip and the reflective target, resulting in changes in light intensity and the output voltage (Figure 72). However, this configuration had a number of disadvantages. The received light intensity was inadequate to observe voltage changes over the entire 3” stroke of the
lift cylinder since most of the light was reflected away from the receiver. In addition, this approach would be sensitive to surface imperfections, and changes in the reflective surface over time. While the light intensity may be increased by using a concave reflective surface to focus the light toward the receiver, this approach would be highly sensitive to misalignment and thus impractical.

![Reflective configuration](image)

**Figure 72: Reflective configuration**

Instead, the transmitter and receiver of the duplex fiber optic cable were split and oriented in transmit-receive configuration, as illustrated in Figure 73. This configuration provided increased light intensity, such that adequate changes in voltage could be seen over the entire 3” range. A focusing lens (60-2748, Rockwell Automation, Inc.) was used at the receiving end in order to further increase the light intensity. The focusing lens was not used at the transmitting end because it focused the light to such an extent that the voltage remained nearly constant over a 3” range, resulting in inadequate sensitivity. Therefore, an MR-compatible series 303 stainless steel sensing tip without lens (FA510, McMaster-Carr, Inc.) was used to mount the fiber at the transmitting end. Using the selected configuration, the voltage ranged between approximately 7 V when the treadmill was fully elevated and 2 V when the treadmill was at its lowest position, resulting in a
difference of approximately 5 V over 3”. The decrease in voltage output with increasing distance between the transmitter and receiver due to reduced light intensity is shown in Figure 81 in Section 7.4.3.

![Diagram of transmit-receive configuration](image)

**Figure 73: Transmit-receive configuration**

The transmit-receive configuration is illustrated in the photo in Figure 74. The MRI-compatible series 300 stainless steel plate used for mounting the receiving (lens) end and the aluminum bracket for mounting the transmitting end were centered with each other along the cylinder. A vertical slot was designed into the aluminum bracket to facilitate alignment. The receiving end was mounted first, and the transmitting end position was adjusted using the vertical slot with the treadmill maximally elevated while simultaneously observing the voltage output in LabVIEW. Once the position corresponding to the maximum voltage output was determined, the sensing tip was fastened to the aluminum bracket.
In the course of system testing, it became apparent that the relationship between the sensor voltage and the inclinometer output appeared to vary at times, and that the variation appeared to be a function of the time since the power supply was turned on. This variation was narrowed down to be due to the elevation sensor, the inclinometer, or the power supply itself. As a result, a drift test was carried out by observing all three components over time using LabVIEW after the power was turned on, and recording the output. Since the DAQ board can only accept inputs up to ±20V, a voltage divider with two 1 kΩ resistors was used to reduce the power supply voltage from 24V down to 12V. The results are displayed in Figure 75, which shows that the elevation sensor has a warm-up time associated with it as illustrated by the decreasing voltage level over time, while the power supply and the inclinometer output are steady. The two notches in the elevation sensor output on the left end of the curve are due to the treadmill being manually raised in two steps such that the voltage level could be monitored at an intermediate elevation.
(the treadmill cannot be raised with the power supply off). A zoomed-in plot clearly illustrating the sensor voltage drift over time is shown in Figure 76. After the point indicated on the curve, the sensor voltage levels were within 1% of the steady state value, which corresponds to 13.8 minutes after the power supply was turned on. This experiment demonstrated that the power supply should be turned on for approximately 15 minutes prior to running the treadmill elevation system. Since it takes at least that long to prepare the patient, set up the MRI room for exercise testing, acquire the resting scans, and obtain the resting ECG’s and blood pressures, this is not anticipated to create any delays.

Figure 75: Drift test showing that the elevation sensor voltage decreases over time after power is switched on until the steady state is reached. The inclinometer and power supply outputs do not vary as a function of time.
Figure 76: The zoomed-in version of Figure 75, clearly illustrating the drift in the elevation sensor voltage, and the time to settle to 1% of the steady-state voltage.

Since the sensor operates within the visible light range (660 nm), an additional test was carried out to observe the effects of turning the room lights on or off. The elevation sensor data was averaged over 180 seconds with the lights on and off, and the difference was measured. This difference was only 0.2%, indicating that ambient light does not affect the sensor output even when the treadmill front cover is removed.

7.4.2 Elevation Control System Design

As discussed in Section 7.1, the elevation control system operates by activating the raise and lower valves to change the stroke of the lift cylinder connected to the leg assembly. At the maximum elevation, the cylinder stroke is at its minimum, resulting in the closest distance between the transmitting and receiving fibers that will be seen during operation and thus the maximum voltage output from the sensor. When the treadmill is
lowered fully, the opposite occurs: the cylinder stroke is at its maximum and the voltage output at its minimum.

The initial elevation control system design was based on activating the raise valve if the elevation was more than 5% below the target and the lower valve if the elevation exceeded the target by more than 5%. However, this caused several oscillations about the target prior to reaching the steady state, thus depleting the accumulator charge volume. Since the accumulator charge needed to be preserved in order to reach higher elevations, the control system was redesigned in such a way as to minimize accumulator charge depletion. To do this, the amount of overshoot needed to be minimized because any excess water in the lift cylinder would be returned to the reservoir, leaving less in the accumulator. As a result, the elevation control system was redesigned using a combination of continuous and pulsed control as described below.

The new elevation control system was designed to achieve the accuracy of ±0.0825° from the target elevation (±7.5% of the stage-to-stage interval of 1.1°). This target accuracy is based on the cylinder stroke only, and does not take into account compliance or bending in the leg assembly. The value of ±7.5% of the stage-to-stage interval was determined experimentally as the minimum acceptable range required to minimize compensation during steady state (in case of, for example, small changes in stroke due to compliance). A value of ±5% of the stage-to-stage interval caused adjustments during steady state which depleted the accumulator charge volume.
Therefore, the selected ±0.0825° error range over which compensation is not performed was chosen to enable precise control while preserving the accumulator charge.

The elevation control system operates by applying continuous control of the raise valve when the error exceeds 0.935° (85% of the stage-to-stage interval of 1.1°), and it switches to pulsed control of the raise valve when the error is between 0.935° and 0.0825°. Continuous control means that the raise valve allows water to flow continuously from the accumulator into the lift cylinder until the error becomes less than 0.935°, with error checks being performed every 250 ms. During pulsed control, the raise valve allows water to flow into the lift cylinder for 60 ms, and then blocks the flow for the subsequent 190 ms (Figure 77). Error checks are performed every 250 ms, and the pulses are applied until the error becomes less than 0.0825°. This type of control can also be referred to as pulse width modulation (PWM) with 4 Hz frequency of pulsing (1 / 0.250 s) and duty cycle of 24% (60 ms / 250 ms x 100%). The angle change, i.e. change in stroke resulting from the application of one pulse is variable depending on the pressure in the accumulator and the subject’s weight. The greater the accumulator pressure, the greater the amount of water which is released into the lift cylinder during the 60 ms interval. The cutoff point between continuous and pulsed control as well as the pulse duration were selected experimentally by trial and error such as to minimize overshoot at the lowest elevations when the accumulator pressure is at its peak and the transition time at higher elevations when the accumulator pressure is at its lowest point. Shorter pulses or a cutoff designed to switch from continuous to pulsed control at greater error magnitudes eliminated any overshoot, but increased the time to reach steady state at higher
elevations. On the other hand, longer pulses or applying continuous control with smaller error magnitudes enabled steady state to be reached more quickly at higher elevations, but caused greater overshoot at lower elevations when the accumulator pressure was at its peak. The number of continuous control intervals and pulses required to undergo transition from stage to stage will be illustrated with an example later in this section.

![Figure 77: Raise pulse timing](image)

The lower valve pulse timing is shown in Figure 78. Once the elevation feedback exceeds the target by more than 0.0825°, the lower valve allows water to flow from the lift cylinder back into the reservoir for 70 ms and then blocks the flow for 180 ms. This can be referred to as PWM with 4 Hz frequency of pulsing and duty cycle of 28%. Although the lower pulse is longer than the raise pulse, less water is released because lowering the treadmill is a slower, gravity-assisted process. The actual amount of water released is a function of the subject’s weight. Therefore, the time to perform adjustments must be balanced with preventing too much water from being released during a single pulse for heavier subjects; the latter could lead to repeat adjustments and drain the accumulator charge volume. Therefore, a 70 ms duration combined with a 28% duty cycle was chosen by trial and error.
The control algorithm is summarized below, where e is defined as the difference between the feedback and the target \( e = \text{elevation feedback} - \text{target elevation} \):

Case 1: \( e \leq -0.935^\circ \)
- Raise valve activated continuously

Case 2: \(-0.935^\circ < e < -0.0825^\circ \)
- Raise valve activated for 60 ms

Case 3: \(-0.0825^\circ < e < 0.0825^\circ \)
- No compensation performed

Case 4: \( e > 0.0825^\circ \)
- Lower valve activated for 70 ms

The LabVIEW implementation of this algorithm is discussed in Section 7.7.4.3.

Although more detailed performance testing will be described in Section 7.8.2, the number of continuous control intervals, raise pulses, and lower pulses applied during stage-to-stage transitions for a 177 lb person are presented in Table 11 to help the reader
gain a better understanding of the system. The data was obtained inside the MRI room at 6 elevations corresponding to 6 stages of the Bruce protocol. It can be observed that lower pulses are applied only during the transition from stage 1 to 2 due to a small overshoot (0.16° above the target of 6.8°, i.e. 2.37%) which is not present at higher stages when the accumulator pressure is lower. This also leads to a longer transition time from stage 1 to 2 than stage 2 to 3. As expected, the number of raise pulses required to transition from stage-to-stage increases at higher stages up to 18 pulses between stages 5 and 6.

Table 11: Number of continuous control intervals, raise pulses, and lower pulses required to transition from stage to stage of the Bruce protocol for a 177 lb subject. The transition time is also shown.

<table>
<thead>
<tr>
<th>Stage transition</th>
<th>Continuous control intervals</th>
<th>Number of raise pulses</th>
<th>Number of lower pulses</th>
<th>Transition time (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 to 2</td>
<td>1</td>
<td>3</td>
<td>5</td>
<td>2.5</td>
</tr>
<tr>
<td>2 to 3</td>
<td>2</td>
<td>2</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>3 to 4</td>
<td>1</td>
<td>7</td>
<td>0</td>
<td>2</td>
</tr>
<tr>
<td>4 to 5</td>
<td>2</td>
<td>11</td>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>5 to 6</td>
<td>2</td>
<td>18</td>
<td>0</td>
<td>5</td>
</tr>
</tbody>
</table>

7.4.3 Calibration

In order to relate the cylinder stroke, measured in the form of voltage output from the elevation sensor, to the angle in degrees, an electronic inclinometer (Spectrotilt™, Spectron Systems Technology Inc., Hauppauge, NY) was used which provides an analog
voltage output as a linear function of the absolute angle with respect to the gravity axis (60 mV/°). The inclinometer was mounted on the treadmill as in Figure 79. The inclinometer specifications include a linear range of 60°, resolution of 0.001°, and repeatability of 0.05°. Since the inclinometer requires a power supply between 9 and 15V, a voltage divider with two 1 kΩ resistors was used to reduce the voltage output from the existing 24 V power supply down to 12 V. The inclinometer voltage was acquired using LabVIEW, and the following conversion was used to convert between the inclinometer readout and the absolute angle:

\[
\text{Angle}(°) = \frac{\text{Inclinometer output (V)} - \text{Zero offset (V)}}{0.06 V_0}
\]  

Equation 9

where zero offset (voltage output at zero angle) = 0.106046 V.

Figure 79: Inclinometer mounted on the treadmill
The inclinometer is a fully signal conditioned electrolytic tilt sensor, and contains an electrolyte sealed within a cavity. The electrolyte conducts between a common, positive, and negative electrode such that when the sensor is level and both electrodes are evenly submerged, a balanced output is produced. When the sensor tilts, this causes an imbalance in the output of the positive and negative electrode resulting from different amounts of electrode surface area being submerged, which is linearly proportional to the angle of tilt. As a result, the inclinometer is able to provide an output relative to the gravity axis. An effect of the fluid-based nature of the inclinometer is sensitivity to rapid motion and acceleration, which can cause fluid “sloshing” inside the cavity. While the sensor performs well under relatively static conditions, this limits the response time of the sensor, and restricts its utility in dynamic applications.

The inclinometer was used to calibrate the elevation sensor by relating its output to an absolute angle. The calibration was performed by simultaneously acquiring the elevation sensor output and the inclinometer output in LabVIEW across a range of angles, and performing a polynomial fit to determine the polynomial coefficients. The original calibration process involved charging the accumulator when the treadmill was at its lowest elevation and then raising the treadmill to its maximum elevation while recording the data in LabVIEW. However, this approach consistently overestimated the inclinometer output at lower elevations. It was subsequently recognized that since the accumulator pressure is the highest at the lowest elevations and becomes depleted as more fluid is released into the lift cylinder, the treadmill was moving at the highest velocity at the lowest elevations, which likely caused “sloshing” of the fluid inside the
inclinometer with inadequate settling time. As a result, the calibration approach was modified to use 250 ms pulses and wait 1 s after each pulse prior to acquiring data in order to allow time for the fluid to settle. Furthermore, the calibration was performed starting at the highest angle and activating the lower valve to progressively lower the treadmill, which is a slower gravity-assisted process independent of the accumulator. This further attenuated the dynamics and eliminated the angle over-estimation.

As previously described, the elevation system is not perfectly rigid due to compliance introduced by air that enters the system, due to hose compliance, and due to mechanical deflection. As a result, instead of performing the calibration with no load, the calibration was performed with a subject of an intermediate weight (177 lbs) standing at the geometric center of the treadmill (midpoint from front to back). The extent of bending, and the control system performance across various weights using this calibration curve, are discussed in Section 7.8.2.

The calibration process was automated in LabVIEW and consists of the following phases with the subject standing at the geometric center of the treadmill:

1. Raise the treadmill fully
2. Simultaneously acquire inclinometer and displacement sensor output every 1.25 seconds while lowering the treadmill (250 ms lower valve pulse plus 1 second pause). The LabVIEW structure is illustrated in Figure 80.
3. Apply a 3rd degree polynomial fit to the inclinometer output vs. elevation sensor output curve, and compute the R-square goodness of fit coefficient.

4. Save the polynomial coefficients \((a_0, a_1, a_2, a_3)\) into a text file titled "Calibration.txt". This text file is subsequently read by the main program, and the coefficients are extracted to convert the elevation sensor feedback into an absolute angle. For example, a displacement sensor output of 4 V would correspond to:

\[
Angle(°) = a_3(4)^3 + a_2(4)^2 + a_1(4) + a_0
\]

Figure 80: LabVIEW program for acquiring the calibration data as the treadmill lowers.

An example of the elevation sensor and inclinometer outputs is shown in Figure 81 as the treadmill lowers. As the cylinder stroke increases, the light intensity and the voltage output decrease due to the greater separation between the transmitter and
receiver. The corresponding 3rd degree polynomial fit of the inclinometer angle as a function of the displacement sensor voltage is shown in Figure 82. A high R-square value of 0.999 indicates a good fit.

Figure 81: Elevation sensor output in volts (white) and inclinometer output in degrees (red) as the treadmill is lowered

Figure 82: Calibration curve showing the inclinometer angle vs. the elevation sensor voltage, with the $R^2$ value displayed
7.5 Failure Modes and Effects Analysis

Since the hydraulic treadmill is a medical device which carries risk of injury to the patient if some of the components were to fail during operation, an extensive failure modes and effects analysis (FMEA) was conducted and categorized into human safety issues, machine safety issues, and non-safety issues which may impact accuracy. The analysis was performed for the following three situations: pretest (charging the accumulator and raising the treadmill to an initial elevation), startup (starting the belt with no PI control), and exercise test. The corresponding analysis is presented in Table 12, Table 13, and Table 14, respectively. The combination of pretest and startup in Table 12 and Table 13 should be able to detect any failure which may have occurred between the previous and current treadmill test, while any new failures which occur during exercise are addressed in Table 14. For each failure mode, its effects and potential causes are presented, as well as the LabVIEW software controls whose implementation will be discussed in Section 7.7.4.4, and the software-independent (non-LabVIEW) safety controls which remain functional even in the case of computer or DAQ failure. All of the listed safety features have been implemented except a water level sensor for the reservoir, which has been selected but not yet installed.
Table 12: Failure modes and effects analysis for the Pretest phase

<table>
<thead>
<tr>
<th>Failure Type</th>
<th>Failure Mode</th>
<th>Effects</th>
<th>Causes</th>
<th>Non-LabVIEW Controls</th>
<th>LabVIEW Controls</th>
</tr>
</thead>
</table>
| P1 Human safety | No elevation feedback | Treadmill will rise up to maximum elevation | • Obstruction in light path between transmitting and receiving fibers  
• Fiber quick disconnects not connected  
• Fiber damage  
• Fiber cables disconnected from sensing tips or elevation sensor  
• No power supplied to elevation sensor  
• Elevation sensor failure | If feedback signal is <1.8V, shut off raise valve and stop motor controller. Warn user. |
| P2 Human safety | Elevation exceeds target angle by >1° during pretest | Treadmill will rise up to maximum elevation | • No power supplied to raise valve  
• Raise valve not plugged in  
• Lower valve does not open | If elevation feedback exceeds target angle by >1°, stop motor controller, shut off raise valve, and lower treadmill. Warn user. |
| P3 Non-safety | Elevation below target by >5% at end of pretest | Treadmill elevation too low, limiting test accuracy | • Lack of power to lower valve  
• Lower valve not plugged in  
• Raise valve does not open | If elevation feedback is more than 5% below target at end of pretest, warn user. |
| P4 Human safety | Flow not diverted from drive system to elevation system | Belt can move unexpectedly during pretest | • Lack of power to accumulator valve  
• Accumulator valve not plugged in | If 2 or more pulses detected during pretest, stop motor controller. Warn user. |
Table 13: Failure modes and effects analysis for the Startup phase

<table>
<thead>
<tr>
<th>Failure Type</th>
<th>Failure Mode</th>
<th>Effects</th>
<th>Causes</th>
<th>Non-LabVIEW Controls</th>
<th>LabVIEW Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1</td>
<td>Non-safety</td>
<td>No speed feedback signal at startup</td>
<td>Electric motor maintains commanded speed. Belt may move at low speed.</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Obstruction in light path between transmitting and receiving fibers</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Fiber quick disconnects not connected</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Fiber damage</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Fiber cables disconnected from sensing tips or speed sensor</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• No power supplied to speed sensor</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Speed sensor failure</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>If no feedback, system will time out in 30 sec and stop</td>
<td></td>
<td></td>
</tr>
<tr>
<td>S2</td>
<td>Machine safety</td>
<td>Zero speed feedback signal at startup</td>
<td>Electric motor maintains speed. Belt is stopped.</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Hose couplers not connected at startup</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Drive shaft coupler fails</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>If no feedback, system will time out in 30 sec and stop</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Failure Type</td>
<td>Failure Mode</td>
<td>Effects</td>
<td>Causes</td>
<td>Non-LabVIEW Controls</td>
<td>LabVIEW Controls</td>
</tr>
<tr>
<td>--------------</td>
<td>--------------</td>
<td>---------</td>
<td>--------</td>
<td>----------------------</td>
<td>-----------------</td>
</tr>
</tbody>
</table>
| T1 Human safety | Loss of speed feedback signal during test | Electric motor and treadmill speed up | • Obstruction in light path between transmitting and receiving fibers  
• Fiber quick disconnects fail  
• Fiber damage  
• Fiber cables disconnect from sensing tips or speed sensor  
• Loss of power to speed sensor  
• Speed sensor failure | 1. Automatic stop if electric motor speed exceeds 75 Hz  
2. Emergency stop button | Stop motor controller if feedback signal is below limits (10%FS or 0.6 mph). During transitions, ensure encoder feedback is between the 2 speeds. |
| T2 Non-safety | Zero speed feedback detected during test | Electric motor speeds up and treadmill stops | • Hose couplers fail  
• Drive shaft coupler fails | 1. Automatic stop if electric motor speed exceeds 75 Hz  
2. Emergency stop button | Stop motor controller if feedback signal is below limits (10%FS or 0.6 mph). During transitions, ensure encoder feedback is between the 2 speeds. |

Continued
### Table 14 Continued

<table>
<thead>
<tr>
<th>Failure Type</th>
<th>Failure Mode</th>
<th>Effects</th>
<th>Causes</th>
<th>Non-LabVIEW Controls</th>
<th>LabVIEW Controls</th>
</tr>
</thead>
</table>
| T3 Human safety | Loss of elevation feedback signal during test | Treadmill will rise to maximum elevation (if accumulator charge allows), which poses a safety concern at high speeds | • Obstruction in light path between transmitting and receiving fibers  
• Fiber quick disconnects fail  
• Fiber damage  
• Fiber cables disconnect from sensing tips or elevation sensor  
• Loss of power to elevation sensor  
• Elevation sensor failure | Emergency stop button | If feedback signal <1.8V, shut off raise valve and stop motor controller |
| T4 Human safety | Raise valve failure | Treadmill will rise to maximum elevation (if accumulator charge allows) | • Loss of power to raise valve | Emergency stop button | If overshoot exceeds limits (10%FS or 1.13°), stop motor controller and lower treadmill. During transitions, ensure sensor feedback is within the 2 elevations. |

Continued
<table>
<thead>
<tr>
<th>Failure Type</th>
<th>Failure Mode</th>
<th>Effects</th>
<th>Causes</th>
<th>Non-LabVIEW Controls</th>
<th>LabVIEW Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td>T5 Human safety</td>
<td>Electrical failure resulting in 5V desired speed input into motor controller</td>
<td>Flywheel speed can increase up to 100 Hz</td>
<td>• Electrical failure such as an arc in wiring between DAQ and motor controller</td>
<td>1. Automatic stop if electric motor speed exceeds 75 Hz 2. Emergency stop button</td>
<td>If speed exceeds 10%FS (0.6 mph) above desired at any speed, stop motor controller. During transitions, ensure encoder feedback is within the 2 limits.</td>
</tr>
<tr>
<td>T6 Machine safety</td>
<td>Slow leak in hose carrying water to hydraulic motor</td>
<td>Treadmill slows down. Feedback control system tries to compensate and pumps more water out. Water may leak within the MRI room, or over treadmill powerpack and other equipment.</td>
<td>Mechanical stress</td>
<td>1. Automatically shut off motor controller if water level switch tripped in order to minimize amount of water spilled 2. Emergency stop button if leak visually detected</td>
<td>1. Implement water level sensor (go from warning to hard stop depending on degree of water level drop) 2. Stop motor controller if feedback signal is more than 10%FS (0.6 mph) below target. During transition, ensure feedback is within the 2 limits.</td>
</tr>
<tr>
<td>Failure Type</td>
<td>Failure Mode</td>
<td>Effects</td>
<td>Causes</td>
<td>Non-LabVIEW Controls</td>
<td>LabVIEW Controls</td>
</tr>
<tr>
<td>--------------</td>
<td>--------------</td>
<td>---------</td>
<td>--------</td>
<td>-----------------------</td>
<td>------------------</td>
</tr>
<tr>
<td>T7 Machine safety</td>
<td>Hose carrying water to hydraulic motor bursts</td>
<td>Treadmill slows down and stops. Feedback control system tries to compensate and pumps more water out. Water discharge may occur within the MRI room, or over treadmill powerpack and other equipment.</td>
<td>Mechanical stress</td>
<td>1. Automatic stop if electric motor speed exceeds 75 Hz 2. Automatically shut off motor controller if water level switch tripped in order to minimize amount of water spilled 3. Emergency stop button if leak visually detected</td>
<td>1. Implement water level sensor (go from warning to hard stop depending on degree of water level drop) 2. Stop motor controller if feedback signal is more than 10%FS (0.6 mph) below target. During transition, ensure feedback is within the 2 limits.</td>
</tr>
<tr>
<td>T8 Machine safety</td>
<td>Failure of hose carrying water back to reservoir (slow leak due to lower pressures)</td>
<td>Water leak within the MRI room or over treadmill powerpack and other equipment. Treadmill speed is not affected.</td>
<td>Mechanical stress</td>
<td>1. Automatically shut off motor controller if water level switch tripped in order to minimize amount of water spilled</td>
<td>1. Implement water level sensor (go from warning to hard stop depending on degree of water level drop)</td>
</tr>
<tr>
<td>T9 Human safety</td>
<td>Loss of LabVIEW START/STOP command</td>
<td>Cannot start or stop treadmill from LabVIEW. Treadmill continues to follow protocol.</td>
<td>Wire connecting DAQ output to motor controller input becomes damaged or disconnected</td>
<td>Emergency stop button</td>
<td></td>
</tr>
<tr>
<td>Failure Type</td>
<td>Failure Mode</td>
<td>Effects</td>
<td>Causes</td>
<td>Non-LabVIEW Controls</td>
<td>LabVIEW Controls</td>
</tr>
<tr>
<td>--------------</td>
<td>----------------------</td>
<td>----------------------------------------------------------------------------------------------------------------</td>
<td>-------------------------------------------------------------------------------------------------------------</td>
<td>------------------------------------------------------------------------------------------</td>
<td>----------------------------------------------------------------------------------</td>
</tr>
<tr>
<td>T10</td>
<td>Human safety</td>
<td>DAQ board failure</td>
<td>Complete loss of LabVIEW control. Motor on and all valves activated (including accumulator). Belt stops and treadmill slowly rises until electric motor speed limit is reached.</td>
<td>DAQ board becomes unplugged or burns out</td>
<td>1. Automatic stop if electric motor speed exceeds 75 Hz 2. Emergency stop button</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>T11</td>
<td>Non-safety (accuracy)</td>
<td>Elevation below limits of accuracy, but not safety limits</td>
<td>Treadmill will lower slowly, which prevents protocol execution</td>
<td>Loss of power to lower valve, lower valve becomes unplugged during test, or raise valve does not open</td>
<td>If elevation feedback is &gt;2%FS (0.226°) below desired over 5 sec, warn user with blinking indicator and record event</td>
</tr>
<tr>
<td>T12</td>
<td>Non-safety (accuracy)</td>
<td>Elevation above limits of accuracy, but not safety limits</td>
<td>Cannot compensate for overshoot in going from stage to stage</td>
<td>Lower valve does not open</td>
<td>If elevation feedback is &gt;2%FS (0.226°) above desired over 5 sec, warn user with blinking indicator and record event</td>
</tr>
</tbody>
</table>

Each failure mode was ranked on the scale of 1 to 5 based on the expected frequency of occurrence and the severity of effects, and the results are presented in Table 15. The graph of severity vs. frequency is displayed in Figure 83, illustrating that none of the failure modes are expected to occur with both high frequency and high severity.
Table 15: Frequency and severity ranking of each failure mode in Table 12, Table 13, and Table 14

<table>
<thead>
<tr>
<th>Failure Mode</th>
<th>Frequency (1=Low, 5=High)</th>
<th>Severity (1=Low, 5=High)</th>
</tr>
</thead>
<tbody>
<tr>
<td>P1</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>P2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>P3</td>
<td>2</td>
<td>1</td>
</tr>
<tr>
<td>P4</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>S1</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>S2</td>
<td>2</td>
<td>1</td>
</tr>
<tr>
<td>T1</td>
<td>2</td>
<td>5</td>
</tr>
<tr>
<td>T2</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>T3</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>T4</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>T5</td>
<td>1</td>
<td>5</td>
</tr>
<tr>
<td>T6</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>T7</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>T8</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>T9</td>
<td>1</td>
<td>4</td>
</tr>
<tr>
<td>T10</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>T11</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>T12</td>
<td>1</td>
<td>1</td>
</tr>
</tbody>
</table>
Figure 83: Plot of failure mode severity vs. frequency, illustrating that none of the failure modes are in the high frequency – high severity quadrant

In addition, a testform was created to test the functionality of the implemented safety features under various failure conditions during pretest, startup, and exercise. This testform is included in Appendix B.

7.6 Implementation of Software-Independent Safety Features

The non-LabVIEW safety features are comprised of two elements: an emergency stop button mounted on the treadmill, and a safety feature to automatically shut off the electric motor if its speed exceeds the maximum speed anticipated during operation. The
emergency stop button implementation will be described first, followed by the speed safety shutoff.

Figure 84: Emergency stop button

The emergency stop button must be mounted on the treadmill inside the MRI room, and as such it must be non-magnetic and have no electrical cables going in or out of the MRI room which can bring in RF noise. Furthermore, it must act independently of the DAQ and the computer since a failure in either must not impact its functionality. The selected emergency stop button (SFS-EBM-01E1, Banner, Inc., Minnesota) is shown mounted on the treadmill frame in Figure 84, and is made entirely of plastic and aluminum. The signal detection is performed using the same sensor model as that used for speed sensing, and the signal transmission is carried out with the same type of plastic fiber optic cables with quick disconnects used in both speed and elevation sensing,
resulting in a more efficient design. The transmitting and receiving fibers were oriented in a through-beam configuration for increased reliability.

The mechanism of operation of the emergency stop button is illustrated in Figure 85. When the button is not pressed, there is a free path for light propagation between the transmitting and receiving fibers, and the sensor will output 24 V. When the button is pressed in an emergency situation, a plastic plate which is attached to the button moves down and obstructs the light path, and the sensor output becomes 0 V. The sensor output is directly wired into one of the motor controller digital inputs (input 4 out of 6), which causes the electric motor to stop whenever a 24 V input is not detected. This configuration is a safety feature in itself, because in the case of fiber optic damage or emergency stop sensor failure, the system will not run. This means that the emergency stop button will be operational whenever the electric motor is running.

![Figure 85: Emergency stop button operation](image)

The timing data for coming to a complete stop when pressing the emergency stop button over the range of speeds and elevations corresponding to 6 stages of the Bruce
protocol was acquired with no subject on the treadmill and with a person weighing 177 lbs. A LabVIEW program was set up to compute the amount of time required for the treadmill to come to a complete stop once the emergency stop is pressed. The stopping point was defined as no flywheel pulses being detected over a 250 ms interval, and the results are presented in Table 16. The time to stop with a subject is shorter at stage 1 due to the additional load, is comparable at stages 2 and 3, and is progressively higher at stages 4, 5, and 6. The likely reason for the increase at higher stages is that the subject is providing an input force to keep moving the belt through their stride (F_∥ in Figure 70). While this reported time is the time for the belt to come to a complete stop, the belt would have slowed down greatly before that point.

Table 16: Time required for the treadmill to come to a complete stop after pressing the emergency stop button across a range of speeds with no subject on the treadmill and a 177 lb person on the treadmill

<table>
<thead>
<tr>
<th>Speed (mph)</th>
<th>Time to stop with no subject (s)</th>
<th>Time to stop with 177 lb subject (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.7</td>
<td>1.985</td>
<td>1.136</td>
</tr>
<tr>
<td>2.5</td>
<td>2.475</td>
<td>2.185</td>
</tr>
<tr>
<td>3.4</td>
<td>3.986</td>
<td>3.835</td>
</tr>
<tr>
<td>4.2</td>
<td>4.486</td>
<td>6.434</td>
</tr>
<tr>
<td>5.0</td>
<td>4.986</td>
<td>8.086</td>
</tr>
<tr>
<td>5.5</td>
<td>5.236</td>
<td>9.635</td>
</tr>
</tbody>
</table>

The second software-independent feature is an automatic shutoff when the electric motor speed exceeds the highest speed expected during operation (above stage 6 of the Bruce protocol). This electric motor speed limit was set to 75 Hz based on the relationship between the flywheel speed and the electric motor speed derived from Figure
67, and a margin to allow the motor to compensate for changes in loading. The motor controller has two digital outputs, which can be set to change state if the electric motor exceeds a specified frequency in Hz. The wiring is schematically illustrated in Figure 86. The 24 V output of the power supply is wired into the “not closed” (N.C.) terminal of digital output 1 while the ground of the power supply is wired into the “not open” (N.O.) terminal of digital output 1. This configuration means that the digital output 1 common will generate a 24 V output while the electric motor speed is below the specified frequency, and when that frequency is exceeded, the output will be 0 V. The digital output 1 common is directly wired into digital input 5, causing the electric motor to automatically stop if the specified frequency is exceeded. The relevant motor controller parameters (365, 380, and 381) are included in the table in Appendix A.

![Wiring Diagram](image)

Figure 86: Wiring illustration for automatically shutting off the motor controller if the electric motor speed exceeds a specified threshold

7.7 LabVIEW Control Program

The goal of the LabVIEW control program was to enable automatic execution of the Bruce treadmill protocol, and to be able to control the treadmill based on any
manually entered speed and elevation up to stage 6 of the Bruce protocol. In addition, the program must incorporate the safety features presented in Section 7.5.

7.7.1 User Interface and Communications

The original LabVIEW user interface is shown in Figure 87. However, during an exercise test, the subject’s ECG is continuously monitored using Cardiosoft software (GE Healthcare, Piscataway, NJ), which prints a predefined number of ECG strips during each stage of the Bruce protocol (typically in 1-minute intervals), including the blood pressures entered by a physician, and uses the information on how long the patient has been on the treadmill to compute the METS (defined in Section 7.3). This software would normally be used to run a GE stress treadmill synchronously with ECG monitoring. Therefore, starting the Bruce protocol in Cardiosoft must occur at the same time as starting the treadmill in LabVIEW. The communications schematic is shown in Figure 88. Due to the limitations in the length of the ECG cable attached to the CAM-14 ECG module on the patient, the Cardiosoft PC must be located within the MRI control room since the waveguide connecting the exam room to the equipment room is positioned near the ceiling and further away from the MRI patient table. The ECG traces are displayed using a shielded in-room monitor. The setup was previously illustrated in Figure 31. The Cardiosoft PC communicates with the keyboard and mouse via a wireless Bluetooth transceiver located inside the MRI room, which is connected to the PC through a waveguide using a shielded USB cable. In order to avoid two separate monitors inside the
MRI room, “Remote Desktop” communication was used to superimpose the LabVIEW window onto the Cardiosoft screen. In order to accomplish this, the original LabVIEW window shown in Figure 87 was reduced to the smaller window shown on the side of the Cardiosoft screen in Figure 89. A blowup of this user interface with a detailed description of the function of each button is shown in Figure 107 in Appendix C.

Figure 87: Original LabVIEW user interface
Figure 88: Layout of communications showing the LabVIEW PC in the equipment room and the Cardiosoft PC in the control room using the same monitor during exercise testing.
7.7.2 Structure of the LabVIEW Control Program

The overall structure of the LabVIEW control program is shown in Figure 90. The program consists of a while loop with three case structures, each linked to a button on the UI: Start/Stop Test, Pretest, and Lower Treadmill. Once the Start/Stop button is pressed,
the program checks for the status of PI control; if PI control status is false (default), the program executes the startup mode described in Section 7.3.4. After successful startup execution, the PI control status is set to true and the main program executes. The Pretest and the Main Program will be described in more detail in the subsequent sections.

Figure 90: Outline of top level of LabVIEW control program

7.7.3 Pretest

The objective of the Pretest loop is to raise the treadmill to stage 1 of the Bruce protocol (or any other commanded angle) while simultaneously charging the accumulator, thus simplifying the Pretest phase to one step. The belt remains stationary throughout the Pretest while the accumulator is charging. The reason for elevating the
treadmill during the Pretest is to eliminate the need to use any of the accumulator charge volume to lift the treadmill to the initial elevation at the start of the test.

The Pretest phase is set up to execute for 8 seconds to ensure that the accumulator is fully charged, but can also be stopped by the user at any time. The commands involved in the Pretest stage include turning off PI control such that the control system does not try to compensate for zero belt speed while the accumulator is charging, starting the electric motor, and charging the accumulator by activating the accumulator valve. During the Pretest, a simplified elevation control is applied such that the raise valve is activated if the feedback is more than 5% below desired and the lower valve is activated if the feedback is more than 5% above desired. The safety checks listed in Table 12 were implemented during the Pretest. In addition, a final accuracy check is performed at the end of the Pretest to determine if the treadmill elevation is more than 5% below the target angle; in the case of stage 1 of the Bruce protocol, this corresponds to 0.285°. The user is alerted to any potential failures or warnings using dialog boxes corresponding to each type of error. The Pretest panel is shown in Figure 91, which can be used to change the desired angle, the allowable error, the duration of the Pretest phase, and the desired safety limits. In addition, the user can observe the changes in sensor voltage and the angle during the Pretest. The desired angle is the most likely parameter to be altered, which would occur when the user does not wish to follow the standard Bruce protocol. Another example where these parameters may change is when using an accumulator with larger capacity, in which case it may not be necessary to elevate to stage 1 during the Pretest,
and the desired angle may be zero. A different accumulator may also require a different charge time.

Figure 91: Pretest panel of the user interface

7.7.4 Main Program

The structure of the main program is displayed in Figure 92.

Figure 92: Main Program
7.7.4.1 Loop Timing

The timing of the main program is controlled by a while loop sequenced immediately prior to acquiring the pulse counter output (Figure 108 in Appendix C). This while loop has a timer which ensures at least 250 ms have elapsed since the previous iteration of the main loop (iteration time t_i). This allows at least 13 pulses on the flywheel to be counted during this time interval even at the lowest speed (1.7 mph) using the target with 5 PPR (11 Hz x 5 PPR / 0.25 s = 13.75 pulses). The rest of the loop, including a 60 ms raise pulse or a 70 ms lower pulse applied if necessary, occur within this time interval of 250 ms.

The Bruce protocol is automatically executed by comparing the time since the start of the test to a set of six cases corresponding to the six stages of the Bruce protocol. For example, if the time in milliseconds is between 1 and 180,000, i.e. the first 3 minutes of the protocol (stage 1), the commanded speed is assigned 1.7 mph and the commanded elevation is assigned 5.7°. These values are then passed through a case structure which checks whether the user has pressed the manual mode command, which can occur at any point during the test. If not, the speed and elevation continue unaltered for the remainder of the iteration, while the manual mode controls on the UI are set to display the current Bruce protocol commanded speed and elevation by default. If the manual mode button is pressed, the speed and elevation are initialized by the latest values from the Bruce protocol, and then replaced by the user entries in the manual mode controls on the UI. This is illustrated in Figure 109 in Appendix C. The test is set to stop automatically if the
end of stage 6 is reached, unless the manual mode button is pressed, in which case it may continue indefinitely until the user stops the test.

### 7.7.4.2 Speed Feedback

The pulse counter is continuously operating independent of software timing, and is used to compute the speed feedback signal (refer to Figure 110 in Appendix C). This is accomplished by finding the difference between the current pulse count and the pulse count at the previous iteration of the main loop, which is converted to belt speed as follows:

\[
Belt \text{ speed}(mph) = \frac{(\text{Pulse count}(t) - \text{Pulse count}(t - t_i)) \times 6.464}{t_i \times 5\text{PPR}}
\]

Equation 10

The conversion factor of 6.464 between flywheel speed in Hz and belt speed in mph was previously derived in Section 7.3. The user interface panel related to the speed feedback is shown in Figure 93. The primary application for this panel would be to visually monitor the speed feedback relative to the desired speed.
7.7.4.3 Elevation Control

The LabVIEW UI panel pertaining to elevation control is displayed in Figure 94. This panel can be used to monitor the elevation feedback relative to the desired elevation, to change the directory of the calibration file, and to view the calibration coefficients. In addition, it can be used to change the cutoffs between continuous and pulsed control, the acceptable error limits, and the duration of the “raise” and “lower” pulses. “Continuous” elevation control means that the raise valve is continuously activated, with checks for whether the error still exceeds 0.935° or 85% of the stage-to-stage interval (Section 7.4.2) being performed every 250 ms (once per each iteration of the main loop). “Pulsed”
control means that a single raise pulse of 60 ms duration is applied once per iteration of the main loop within the 250 ms interval (Figure 77). Similarly, a single lower pulse of 70 ms duration is applied within the 250 ms interval (Figure 78) when the elevation feedback exceeds the target by more than 0.0825°. These steps are illustrated schematically in Figure 95. The raise control implementation is shown in Appendix C and illustrated below in the flowchart in Figure 96. Figure 111 in Appendix C shows the case where no elevation control is needed, Figure 112 the case where the pulsed error and the continuous error are both exceeded, and Figure 113 the case where the pulsed error is exceeded but the continuous error is not. The lower pulse implementation is identical to the raise pulse, except that the duration is 70 ms instead of 60 ms, and is not shown here.

Figure 94: LabVIEW elevation panel
7.7.4.4 Speed and Elevation Safety Control Implementation

The speed safety feature illustration is shown in Figure 97, where $E$ is equal to 10% of full scale or 0.6 mph above or below the target. During the transitions, the limits
are set to Target 1 - $E$ and Target 2 + $E$ when Target 2 > Target 1. Conversely, the limits are set to Target 1 + $E$ and Target 2 - $E$ during the transitions when Target 2 < Target 1.

The speed failure mode is implemented according to the following algorithm ($t$ is the current time in seconds):

Case 1: Target ($t$) = Target ($t$ – 3 sec), i.e. Target 2 = Target 1 (not in transition)

Upper Limit = Target 2 + 0.6 mph

Lower Limit = Target 2 – 0.6 mph

Case 2: Target ($t$) > Target ($t$ – 3 sec), i.e. Target 2 > Target 1 (increasing speed)

Upper Limit = Target 2 + 0.6 mph

Lower Limit = Target 1 – 0.6 mph

Case 3: Target ($t$) < Target ($t$ – 3 sec), i.e. Target 2 < Target 1 (decreasing speed)

Upper Limit = Target 1 + 0.6 mph

Lower Limit = Target 2 – 0.6 mph

Figure 97: Implementation of speed safety limits, including transitions
The LabVIEW computation of the speed limits, and the checks to determine if the speed feedback is above or below the limits, are shown in Figure 114 in Appendix C. The speed feedback must be within both limits; otherwise, a stop command is immediately sent to the motor controller (Figure 115 in Appendix C), a dialog box indicating the error is displayed, and the Start/Stop button is automatically turned off. The elevation control sequence is bypassed, and the main while loop is stopped.

The same principle illustrated in Figure 97 is used to implement the elevation safety features, the difference being that the transition mode lasts for 5 seconds, and E corresponds to 10% of elevation full scale or 1.13°. The algorithm is described below:

Case 1: Target (t) = Target (t – 5 sec), i.e. Target 2 = Target 1 (not in transition)
   \[
   \text{Upper Limit} = \text{Target } 2 + 1.13^\circ \\
   \text{Lower Limit} = \text{Target } 2 - 1.13^\circ 
   \]

Case 2: Target (t) > Target (t – 5 sec), i.e. Target 2 > Target 1 (increasing elevation)
   \[
   \text{Upper Limit} = \text{Target } 2 + 1.13^\circ \\
   \text{Lower Limit} = \text{Target } 1 - 1.13^\circ 
   \]

Case 3: Target (t) < Target (t – 5 sec), i.e. Target 2 < Target 1 (decreasing elevation)
   \[
   \text{Upper Limit} = \text{Target } 1 + 1.13^\circ \\
   \text{Lower Limit} = \text{Target } 2 - 1.13^\circ 
   \]

All safety features can be turned on or off by checking or un-checking the boxes on the Safety panel (Figure 98). When the program is first opened, all safety features are
set to be on by default. The purpose of allowing the safety features to be temporarily disabled is when troubleshooting the system or implementing new features. In addition, this panel can be used to change the safety limits if desired.

Figure 98: Safety features panel

7.8 Control System Performance Testing

The aim of the treadmill performance testing was to evaluate the speed and elevation performance outside and inside the MRI room, as well as determine whether any differences are observed when the treadmill operates within the magnetic field. The potential sources of magnetic field effects are eddy currents induced in the flywheel, or
eddy currents induced in the treadmill frame as it rises or lowers within the magnetic field. The treadmill board is made of wood, and is thus not expected to contribute to these effects. The treadmill was positioned parallel to the MRI machine, with the front sitting immediately adjacent to the magnet (Figure 31), which represents the worst-case scenario in terms of the magnetic field strength the power and sensing components may be exposed to during clinical operation. Since the front of the treadmill is located outside of the bore and below the isocenter even at the closest proximity to the magnet, the eddy current effects were not anticipated to adversely affect the control system performance.

7.8.1 Speed Performance Testing

The speed performance was tested using two different tools. The first was a plastic 5th wheel device (Figure 99), which was placed in contact with the moving belt. One full revolution of the device corresponded to 1 ft of belt movement. The second tool was a pulse tachometer, which counted strips of reflective tape placed on the belt. Eight reflective strips were spaced evenly across the entire belt length at a distance of 14.2” between strips. Both sets of measurements were performed over 30 second intervals across all stages of the Bruce protocol, from 1.7 mph and 5.7° to 5.5 mph and 11.3°. The start and stop time for each data acquisition interval was determined using a stopwatch. In addition to the range of speeds and elevations, each measurement was also performed across the following weights: unloaded (no person on treadmill), 125 lb, 177 lb, and 203 lb.
The measurements were first acquired outside of the MRI room using both the 5th wheel and the pulse tachometer. Inside the MRI room, the first step was to validate the electrical pulse tachometer using the fully MRI-compatible plastic 5th wheel device under the unloaded condition (no person on the treadmill). Subsequently, only the pulse tachometer, which is more user-friendly and less subject to operator error than the 5th wheel device, was used for data acquisition across the various stages and weights inside of the MRI room.

The conversion between the 5th wheel readout (F) and the belt speed (\(V_B\)) was as follows:

\[
V_B (mph) = \frac{F (ft)}{t (s)} \times \frac{1 (mile)}{5280 (ft)} \times \frac{3600 (s)}{1 (hr)}
\]

Equation 11

For \(t = 30\) sec, this reduces to:

\[
V_B (mph) = F \times 0.0227272727
\]

Equation 12
The conversion between the pulse count (P) and the belt speed (V_B) was as follows:

\[
V_B (mph) = \frac{P(pulses)}{t(s)} \times \frac{1(rev)}{n(pulses)} \times \frac{L(ft)}{1(rev)} \times \frac{1(mile)}{5280(ft)} \times \frac{3600(s)}{1(hr)},
\]

Equation 13

where

Belt length (L) = 9.5 ft

Number of pulses per revolution (n) = 8.

For t = 30 sec, this reduces to:

\[
V_B (mph) = P \times 0.02698863636
\]

Equation 14

Table 17 shows the average speed data obtained over 30 seconds under the unloaded condition outside and inside of the MRI room, which was acquired using the pulse tachometer and the 5th wheel. The maximum deviation between the pulse tachometer and the target speed outside the MRI room was 1.60%, and the maximum deviation between the 5th wheel and the target speed was 1.00%. Inside the MRI room, the maximum deviation between the pulse tachometer and the target was 1.60% and the maximum deviation between the 5th wheel and the target was 1.23%, indicating close agreement in both cases. Therefore, it was shown that the electrical pulse tachometer could be used reliably inside the MRI room.
Table 17: Average speed data outside and inside of the MRI room acquired over a 30-second interval across all Bruce protocol stages with no person on the treadmill. The data was acquired using a pulse tachometer and a 5th wheel device.

<table>
<thead>
<tr>
<th>OUTSIDE MRI ROOM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stage</td>
</tr>
<tr>
<td>-------</td>
</tr>
<tr>
<td>1</td>
</tr>
<tr>
<td>2</td>
</tr>
<tr>
<td>3</td>
</tr>
<tr>
<td>4</td>
</tr>
<tr>
<td>5</td>
</tr>
<tr>
<td>6</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>INSIDE MRI ROOM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stage</td>
</tr>
<tr>
<td>-------</td>
</tr>
<tr>
<td>1</td>
</tr>
<tr>
<td>2</td>
</tr>
<tr>
<td>3</td>
</tr>
<tr>
<td>4</td>
</tr>
<tr>
<td>5</td>
</tr>
<tr>
<td>6</td>
</tr>
</tbody>
</table>

Table 18 represents the average speed data outside and inside of the MRI room, acquired over 30 seconds with a 125 lb person, a 177 lb person, and a 203 lb person running on the treadmill through each stage of the Bruce protocol. The raw data, consisting of the pulse count and the 5th wheel measurement in feet, is shown in Appendix D. The greatest error among all weights both inside and outside of the MRI room measured with the pulse tachometer was 1.60%. The 5th wheel measurement outside of the MRI room resulted in the greatest error of 1.87%. Therefore, close agreement with the target speed was obtained in each case, meeting the design goal of maintaining steady state error within 2% of the target speed across a range of weights.
The mean and standard deviation of the combined speed error for all subjects outside and inside the MRI room obtained using the pulse tachometer is summarized in Table 19.

Table 18: Average speed data outside and inside of the MRI room acquired over a 30-second interval across all Bruce protocol stages with a 125 lb, a 177 lb, and a 203 lb person running on treadmill.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target Speed (mph)</th>
<th>Pulse Tach (mph)</th>
<th>Difference from Target (%)</th>
<th>5th Wheel (mph)</th>
<th>Difference from Target (%)</th>
<th>Pulse Tach (mph)</th>
<th>Difference from Target (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>125 LBS</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>1.7</td>
<td>1.73</td>
<td>1.60</td>
<td>1.70</td>
<td>0.27</td>
<td>1.70</td>
<td>0.02</td>
</tr>
<tr>
<td>2</td>
<td>2.5</td>
<td>2.51</td>
<td>0.40</td>
<td>2.49</td>
<td>-0.45</td>
<td>2.48</td>
<td>-0.68</td>
</tr>
<tr>
<td>3</td>
<td>3.4</td>
<td>3.40</td>
<td>0.02</td>
<td>3.38</td>
<td>-0.74</td>
<td>3.43</td>
<td>0.81</td>
</tr>
<tr>
<td>4</td>
<td>4.2</td>
<td>4.21</td>
<td>0.24</td>
<td>4.15</td>
<td>-1.24</td>
<td>4.16</td>
<td>-1.04</td>
</tr>
<tr>
<td>5</td>
<td>5.5</td>
<td>5.51</td>
<td>0.10</td>
<td>5.47</td>
<td>-0.50</td>
<td>5.48</td>
<td>-0.39</td>
</tr>
<tr>
<td>177 LBS</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>1.7</td>
<td>1.70</td>
<td>0.02</td>
<td>1.71</td>
<td>0.40</td>
<td>1.71</td>
<td>1.60</td>
</tr>
<tr>
<td>2</td>
<td>2.5</td>
<td>2.51</td>
<td>0.40</td>
<td>2.50</td>
<td>-0.09</td>
<td>2.48</td>
<td>-0.68</td>
</tr>
<tr>
<td>3</td>
<td>3.4</td>
<td>3.37</td>
<td>-0.78</td>
<td>3.41</td>
<td>0.20</td>
<td>3.40</td>
<td>0.02</td>
</tr>
<tr>
<td>4</td>
<td>4.2</td>
<td>4.16</td>
<td>-1.04</td>
<td>4.15</td>
<td>-1.30</td>
<td>4.21</td>
<td>0.24</td>
</tr>
<tr>
<td>5</td>
<td>5.5</td>
<td>5.45</td>
<td>-0.88</td>
<td>5.42</td>
<td>-1.53</td>
<td>5.48</td>
<td>-0.39</td>
</tr>
<tr>
<td>203 LBS</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>1.7</td>
<td>1.73</td>
<td>1.60</td>
<td>1.73</td>
<td>1.87</td>
<td>1.73</td>
<td>1.60</td>
</tr>
<tr>
<td>2</td>
<td>2.5</td>
<td>2.51</td>
<td>0.40</td>
<td>2.49</td>
<td>-0.55</td>
<td>2.48</td>
<td>-0.68</td>
</tr>
<tr>
<td>3</td>
<td>3.4</td>
<td>3.40</td>
<td>0.02</td>
<td>3.37</td>
<td>-1.00</td>
<td>3.37</td>
<td>-0.78</td>
</tr>
<tr>
<td>4</td>
<td>4.2</td>
<td>4.16</td>
<td>-1.04</td>
<td>4.16</td>
<td>-0.87</td>
<td>4.16</td>
<td>-1.04</td>
</tr>
<tr>
<td>5</td>
<td>5.5</td>
<td>5.02</td>
<td>0.40</td>
<td>5.03</td>
<td>0.64</td>
<td>4.94</td>
<td>-1.22</td>
</tr>
<tr>
<td>6</td>
<td>5.5</td>
<td>5.48</td>
<td>-0.39</td>
<td>5.43</td>
<td>-1.19</td>
<td>5.48</td>
<td>-0.39</td>
</tr>
</tbody>
</table>
Table 19: Combined speed error for all subjects (125 lb, 177 lb, 203 lb) outside and inside the MRI room obtained using the pulse tachometer

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target Speed (mph)</th>
<th>Outside Average Error (%)</th>
<th>Outside Stdev Error (%)</th>
<th>Inside Average Error (%)</th>
<th>Inside Stdev Error (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.7</td>
<td>0.85</td>
<td>0.89</td>
<td>1.08</td>
<td>0.92</td>
</tr>
<tr>
<td>2</td>
<td>2.5</td>
<td>0.36</td>
<td>0.24</td>
<td>0.68</td>
<td>0.00</td>
</tr>
<tr>
<td>3</td>
<td>3.4</td>
<td>0.65</td>
<td>0.41</td>
<td>0.53</td>
<td>0.45</td>
</tr>
<tr>
<td>4</td>
<td>4.2</td>
<td>1.14</td>
<td>0.24</td>
<td>0.78</td>
<td>0.46</td>
</tr>
<tr>
<td>5</td>
<td>5</td>
<td>0.82</td>
<td>0.16</td>
<td>0.68</td>
<td>0.54</td>
</tr>
<tr>
<td>6</td>
<td>5.5</td>
<td>1.07</td>
<td>0.53</td>
<td>0.39</td>
<td>0.00</td>
</tr>
</tbody>
</table>

The error with the pulse tachometer at any weight appeared to be greatest at stage 1 (1.60%); however, this is likely due to the fact that the total pulse count is lowest at the slowest speed, and sensitivity to error is the highest. For example, counting one less pulse (63 instead of 64; Appendix D) would have reduced the error from 1.60% to 0.0167%. The added pulse may have been due to the synchronization between starting and stopping the stopwatch and the pulse tachometer. However, this small error was within the design goal, and corresponded to only 0.0272 mph at the stage 1 speed of 1.7 mph. In addition, this speed error corresponds to a METS error of only 1.26%, which is small compared to a 52% METS increase from stage 1 to stage 2. If the maximum error is considered in terms of absolute magnitude rather than a percentage, the maximum error was 0.0611 mph at stage 5 (1.22%). This corresponded to a METS change of only 1.15%.

The difference between the average speed obtained inside and outside of the MRI room using the pulse tachometer is displayed in Table 20 across a range of weights (0, 125 lb, 177 lb, and 203 lb), indicating close agreement. The greatest difference was 1.61% with a 203 lb person at stage 5, while the mean absolute difference was 0.76 ±
0.56%. Therefore, the speed data outside and inside the MRI room were in close agreement, indicating that the magnetic field did not induce eddy currents in the flywheel which the control system could not compensate for.

Table 20: Differences in average speed obtained inside and outside the MRI room using the pulse tachometer at each stage of the Bruce protocol and across a range of weights

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target Speed (mph)</th>
<th>No Load (%)</th>
<th>125 Lbs (%)</th>
<th>177 Lbs (%)</th>
<th>203 Lbs (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.7</td>
<td>0.00</td>
<td>-1.56</td>
<td>1.59</td>
<td>0.00</td>
</tr>
<tr>
<td>2</td>
<td>2.5</td>
<td>0.00</td>
<td>-1.08</td>
<td>-1.08</td>
<td>-1.08</td>
</tr>
<tr>
<td>3</td>
<td>3.4</td>
<td>0.00</td>
<td>0.79</td>
<td>0.80</td>
<td>-0.79</td>
</tr>
<tr>
<td>4</td>
<td>4.2</td>
<td>-0.65</td>
<td>-1.28</td>
<td>1.30</td>
<td>0.00</td>
</tr>
<tr>
<td>5</td>
<td>5</td>
<td>-0.53</td>
<td>-0.54</td>
<td>1.09</td>
<td>-1.61</td>
</tr>
<tr>
<td>6</td>
<td>5.5</td>
<td>-1.46</td>
<td>-0.49</td>
<td>0.50</td>
<td>0.00</td>
</tr>
</tbody>
</table>

7.8.2 Elevation Performance Testing

The elevation control system was validated using the inclinometer to ensure that the target elevations were met over the range of angles corresponding to 6 stages of the Bruce protocol. However, the first step was to validate the accuracy of the inclinometer itself outside and inside of the MRI room, which was conducted using an aluminum 16” x 24” carpenter’s square with 1/16” divisions. Using a LabVIEW program which enabled manual control of the raise and lower valves either in continuous mode or using 100 ms pulses, the treadmill elevation was adjusted until the inclinometer output matched the 6 stages of the Bruce protocol. The carpenter’s square was used to measure the height of the front of the treadmill (H) at each angle, which is illustrated in Figure 100. This
measurement was combined with the height of the pivot \((h = 1.625"\)) and the length of the frame \((L = 62"\)) to obtain the measured angle \((\theta)\) as follows:

\[
\theta = \sin^{-1}\left( \frac{H - h}{L} \right) \times \frac{180^\circ}{\pi}
\]

Equation 15

Figure 100: Validation of inclinometer accuracy using an aluminum carpenter’s square

The results are displayed in Table 21, showing that the agreement between the inclinometer readout and the angle determined through trigonometry using the carpenter’s square measurements was within 0.1° inside or outside of the MRI room at any stage of the Bruce protocol. Therefore, the inclinometer could be used to validate the accuracy of the elevation control system.
Table 21: Comparison between the inclinometer readout and the corresponding angle determined through trigonometry using the carpenter’s square measurements outside and inside the MRI room, showing agreement within 0.1°

<table>
<thead>
<tr>
<th>Stage</th>
<th>Inclinometer (deg)</th>
<th>H (in)</th>
<th>Measured (deg)</th>
<th>Difference from Inclinometer (deg)</th>
<th>H (in)</th>
<th>Measured (deg)</th>
<th>Difference from Inclinometer (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>7.875</td>
<td>5.79</td>
<td>0.09</td>
<td>7.78125</td>
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<tr>
<td>2</td>
<td>6.8</td>
<td>9</td>
<td>6.83</td>
<td>0.03</td>
<td>9</td>
<td>6.83</td>
<td>0.03</td>
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<td>0.00</td>
</tr>
<tr>
<td>4</td>
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<td>11.375</td>
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<td>-0.05</td>
<td>11.375</td>
<td>9.05</td>
<td>-0.05</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>12.5625</td>
<td>10.16</td>
<td>-0.04</td>
<td>12.5</td>
<td>10.10</td>
<td>-0.10</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>13.8125</td>
<td>11.34</td>
<td>0.04</td>
<td>13.6875</td>
<td>11.22</td>
<td>-0.08</td>
</tr>
</tbody>
</table>

The accuracy of the elevation control system was evaluated under static conditions by executing an abbreviated Bruce protocol with 20 seconds per stage across a range of weights. During the test, the subjects stood at the midpoint of the frame of the treadmill, analogous to the calibration run. The steady state elevation sensor feedback and the inclinometer readout acquired outside and inside of the MRI room at each stage are presented in Table 22 for a 125 lb person, Table 23 for a 177 lb person, Table 24 for a 203 lb person, and Table 25 for a 177 lb person carrying a 67 lb backpack for a total weight of 244 lb. The maximum difference observed both outside and inside of the MRI room between the average elevation sensor feedback and the target elevation was 0.07°, which occurred with the 203 lb and 244 lb weights at stage 6. The maximum difference between the average inclinometer output and the target elevation was -0.23° with the 244 lb weight at stage 4. The latter corresponds to an error of 2.5%, which is within the design goal of 3% and corresponds to a METS error of only 1.69%. This is significantly lower than the METS difference of 33% between stage 3 and stage 4. If elevation error percentages are considered instead of the absolute errors, the maximum difference
occurred with the 125 lb subject at stage 2 (0.19° or 2.83%). This was also within the
design target of 3%, and corresponded to a METS change of only 1.65%, which is much
lower than the 44% change in METS between stage 2 and stage 3. The average and the
standard deviation of the elevation error relative to the inclinometer across all weights
outside and inside the MRI room are presented in Table 26.

Table 22: The steady state elevation sensor feedback and the inclinometer readout
acquired outside and inside of the MRI room with a 125 lb subject standing on the frame
of the treadmill. The target elevations corresponded to six stages of the Bruce protocol at
20 seconds per stage.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.70</td>
<td>0.00</td>
<td>5.83</td>
<td>0.13</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.84</td>
<td>0.04</td>
<td>6.99</td>
<td>0.19</td>
<td>2.75</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>8.03</td>
<td>0.03</td>
<td>8.08</td>
<td>0.08</td>
<td>2.25</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.04</td>
<td>-0.06</td>
<td>9.03</td>
<td>-0.07</td>
<td>2.25</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>10.16</td>
<td>-0.04</td>
<td>10.26</td>
<td>0.06</td>
<td>3.00</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.24</td>
<td>-0.06</td>
<td>11.29</td>
<td>-0.01</td>
<td>4.50</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.70</td>
<td>0.00</td>
<td>5.75</td>
<td>0.05</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.82</td>
<td>0.02</td>
<td>6.90</td>
<td>0.10</td>
<td>2.25</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>8.04</td>
<td>0.04</td>
<td>8.01</td>
<td>0.01</td>
<td>2.00</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.05</td>
<td>-0.05</td>
<td>8.98</td>
<td>-0.12</td>
<td>2.25</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>10.14</td>
<td>-0.06</td>
<td>10.22</td>
<td>0.02</td>
<td>2.75</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.25</td>
<td>-0.05</td>
<td>11.30</td>
<td>0.00</td>
<td>4.75</td>
</tr>
</tbody>
</table>

Table 23: The steady state elevation sensor feedback and the inclinometer readout
acquired outside and inside of the MRI room with a 177 lb subject standing on the frame
of the treadmill. The target elevations corresponded to six stages of the Bruce protocol at 20 seconds per stage.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.69</td>
<td>-0.01</td>
<td>5.80</td>
<td>0.10</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.84</td>
<td>0.04</td>
<td>6.96</td>
<td>0.16</td>
<td>3.50</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>8.04</td>
<td>0.04</td>
<td>8.02</td>
<td>0.02</td>
<td>2.25</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
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<td>9.01</td>
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<td>2.75</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>10.17</td>
<td>-0.03</td>
<td>10.24</td>
<td>0.04</td>
<td>3.75</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.24</td>
<td>-0.06</td>
<td>11.28</td>
<td>-0.02</td>
<td>5.50</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
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<td>0.01</td>
<td>5.73</td>
<td>0.03</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.81</td>
<td>0.01</td>
<td>6.85</td>
<td>0.05</td>
<td>2.50</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>8.05</td>
<td>0.05</td>
<td>7.96</td>
<td>-0.04</td>
<td>1.00</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.04</td>
<td>-0.06</td>
<td>8.91</td>
<td>-0.19</td>
<td>2.00</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>10.16</td>
<td>-0.04</td>
<td>10.21</td>
<td>0.01</td>
<td>3.25</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.25</td>
<td>-0.05</td>
<td>11.28</td>
<td>-0.02</td>
<td>5.00</td>
</tr>
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</table>
Table 24: The steady state elevation sensor feedback and the inclinometer readout acquired outside and inside of the MRI room with a 203 lb subject standing on the frame of the treadmill. The target elevations corresponded to six stages of the Bruce protocol at 20 seconds per stage.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>OUTSIDE MRI ROOM</td>
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<td></td>
<td></td>
</tr>
<tr>
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<td>5.81</td>
<td>0.11</td>
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<td>0.03</td>
<td>6.93</td>
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</tr>
<tr>
<td>3</td>
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<td>0.05</td>
<td>8.00</td>
<td>0.00</td>
<td>1.25</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.06</td>
<td>-0.04</td>
<td>8.98</td>
<td>-0.12</td>
<td>2.50</td>
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<tr>
<td>5</td>
<td>10.2</td>
<td>10.16</td>
<td>-0.04</td>
<td>10.23</td>
<td>0.03</td>
<td>4.50</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.23</td>
<td>-0.07</td>
<td>11.28</td>
<td>-0.02</td>
<td>8.50</td>
</tr>
<tr>
<td>INSIDE MRI ROOM</td>
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<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.73</td>
<td>0.03</td>
<td>5.73</td>
<td>0.03</td>
<td>N/A</td>
</tr>
<tr>
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<td>6.80</td>
<td>0.00</td>
<td>6.84</td>
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<td>2.25</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>8.05</td>
<td>0.05</td>
<td>7.92</td>
<td>-0.08</td>
<td>1.25</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.09</td>
<td>-0.01</td>
<td>8.97</td>
<td>-0.13</td>
<td>2.25</td>
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<td>-0.05</td>
<td>10.18</td>
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<td>3.50</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
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<td>-0.05</td>
<td>11.27</td>
<td>-0.03</td>
<td>6.75</td>
</tr>
</tbody>
</table>
Table 25: The steady state elevation sensor feedback and the inclinometer readout acquired outside and inside of the MRI room with a 177 lb subject carrying a 67 lb backpack (total 244 lbs) standing on the frame of the treadmill. The target elevations corresponded to six stages of the Bruce protocol at 20 seconds per stage.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.69</td>
<td>-0.01</td>
<td>5.75</td>
<td>0.05</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.82</td>
<td>0.02</td>
<td>6.90</td>
<td>0.10</td>
<td>2.00</td>
</tr>
<tr>
<td>3</td>
<td>8.0</td>
<td>8.04</td>
<td>0.04</td>
<td>7.75</td>
<td>-0.01</td>
<td>2.00</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>9.06</td>
<td>-0.04</td>
<td>8.95</td>
<td>-0.15</td>
<td>3.00</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>10.16</td>
<td>-0.04</td>
<td>10.00</td>
<td>-0.04</td>
<td>5.00</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.25</td>
<td>-0.06</td>
<td>11.25</td>
<td>-0.06</td>
<td>8.75</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>Elevation Sensor (deg)</th>
<th>Difference from Target (deg)</th>
<th>Inclinometer (deg)</th>
<th>Difference from Target (deg)</th>
<th>Transition (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.7</td>
<td>5.70</td>
<td>0.00</td>
<td>5.64</td>
<td>-0.06</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>6.83</td>
<td>0.03</td>
<td>6.82</td>
<td>0.02</td>
<td>2.25</td>
</tr>
<tr>
<td>3</td>
<td>8.0</td>
<td>8.06</td>
<td>0.06</td>
<td>7.92</td>
<td>-0.08</td>
<td>1.25</td>
</tr>
<tr>
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<td>9.1</td>
<td>9.04</td>
<td>-0.06</td>
<td>8.87</td>
<td>-0.23</td>
<td>3.00</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
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<td>-0.07</td>
<td>10.12</td>
<td>-0.06</td>
<td>5.75</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>11.23</td>
<td>-0.07</td>
<td>11.22</td>
<td>-0.08</td>
<td>10.00</td>
</tr>
</tbody>
</table>

Table 26: Combined elevation error across all weights (125 lb, 177 lb, 203 lb, 244 lb) obtained outside and inside the MRI room relative to the inclinometer.
The mean absolute difference between the sensor feedback and the target across all stages and weights was 0.037 ± 0.018° outside the MRI room and 0.038 ± 0.022° inside the MRI room. The average difference between the inclinometer output and the target was 0.074 ± 0.056° outside of the MRI room, and 0.063 ± 0.057° inside of the MRI room. This indicates a close agreement with the target for both the elevation sensor and the inclinometer inside and outside of the MRI room across a range of weights between 125 lbs and 244 lbs.

The above tables also display the transition time between Bruce protocol stages required for the elevation sensor feedback to reach the steady state level of ±0.0825° from the target elevation (7.5% of the stage-to-stage interval of 1.1°), which the control system is designed to achieve. The transition time at stage 1 is not displayed since the treadmill is elevated to stage 1 during the pretest while the accumulator is being charged, and the treadmill starts at this level at the beginning of the test. The transition time is a function of the accumulator charge, which affects the amount of water released during a given time interval corresponding to a raise or lower pulse. As a result, the transition time generally increases at higher stages when the accumulator is at lower pressure and less water is released during each pulse. While this trend is generally true, the transition time at stage 2 is higher than that at stage 3 due to a small overshoot (approximately 2.4% with the 125 lb subject and 3.6% with the 244 lb subject) at stage 2 when the accumulator charge is at its peak. Finally, the above 4 tables also show that the transition time at higher stages increases with the subject’s weight due to the higher load pressure on the
system induced by the subject. The transition times between stage 5 and stage 6 for the 4 weights are presented in Table 27.

Table 27: Transition time between stage 5 and stage 6 for 4 subject weights obtained outside and inside the MRI room

<table>
<thead>
<tr>
<th>Weight (lbs)</th>
<th>Transition Time from Stage 5 to Stage 6 Outside the MRI room (s)</th>
<th>Transition Time from Stage 5 to Stage 6 Inside the MRI Room (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>125</td>
<td>4.5</td>
<td>4.75</td>
</tr>
<tr>
<td>177</td>
<td>5.5</td>
<td>5.0</td>
</tr>
<tr>
<td>203</td>
<td>8.5</td>
<td>6.75</td>
</tr>
<tr>
<td>244</td>
<td>8.75</td>
<td>10.0</td>
</tr>
</tbody>
</table>

The elevation data acquired inside the MRI room under static conditions with the 125 lb subject over 6 stages at 20 seconds per stage is displayed in Figure 101. The corresponding data acquired with the 244 lb subject is shown in Figure 102. These figures visually illustrate the agreement between both the elevation sensor feedback and the inclinometer readout with the target elevation across a range of weights, as well as the increased transition time at the higher stages and weights.
Figure 101: Elevation target angle, sensor feedback, and inclinometer data for a 125 subject, acquired with the person standing on the frame of the treadmill over 6 stages of the Bruce protocol at 20 seconds per stage

Figure 102: Elevation target angle, sensor feedback, and inclinometer data for a 177 lb subject with a 67 lb backpack (total 244 lbs), acquired with the person standing on the frame of the treadmill over 6 stages of the Bruce protocol at 20 seconds per stage
The differences between the average elevation sensor and inclinometer data acquired outside and inside the MRI room across a range of weights is shown in Table 28 at 6 Bruce protocol stages. The maximum difference in the elevation sensor data was 0.45% with the 203 lb weight at stage 2, while the maximum difference in the average inclinometer readout was 1.95% with the 244 lb weight at stage 1. The mean absolute difference for the elevation sensor was $0.19 \pm 0.12\%$ and for the inclinometer was $0.83 \pm 0.53\%$. These results indicate that any added resistance to motion due to eddy currents in the treadmill frame were compensated by the control system and did not impact accuracy.

Table 28: Differences between elevation sensor and inclinometer data acquired outside and inside the MRI room across a range of weights

<table>
<thead>
<tr>
<th>Stage</th>
<th>Target (deg)</th>
<th>ELEVATION SENSOR</th>
<th></th>
<th>INCLINOMETER</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>125 Lbs (%)</td>
<td>177 Lbs (%)</td>
<td>203 Lbs (%)</td>
<td>244 Lbs (%)</td>
</tr>
<tr>
<td>1</td>
<td>5.7</td>
<td>-0.11</td>
<td>0.28</td>
<td>0.35</td>
<td>0.22</td>
</tr>
<tr>
<td>2</td>
<td>6.8</td>
<td>-0.29</td>
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<td>-0.45</td>
<td>0.08</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>0.10</td>
<td>0.12</td>
<td>0.05</td>
<td>0.15</td>
</tr>
<tr>
<td>4</td>
<td>9.1</td>
<td>0.11</td>
<td>-0.37</td>
<td>0.37</td>
<td>-0.24</td>
</tr>
<tr>
<td>5</td>
<td>10.2</td>
<td>-0.13</td>
<td>-0.08</td>
<td>-0.10</td>
<td>-0.25</td>
</tr>
<tr>
<td>6</td>
<td>11.3</td>
<td>0.06</td>
<td>0.09</td>
<td>0.14</td>
<td>-0.11</td>
</tr>
</tbody>
</table>

Compliance in the system causes some bounce in the treadmill during walking and running, which is primarily due to air (compressible fluid) being present in the system instead of just water (incompressible fluid), and due to the compliance of the hoses. Another possible reason is bending in the aluminum legs of the treadmill. The inclinometer could not be used to measure the extent of the bounce since it caused significant noise in the inclinometer output. As discussed in Section 7.4.3, this is likely
due to the fact that the inclinometer is an electrolytic tilt sensor, such that sudden motion
could cause “sloshing” of the fluid inside the inclinometer and greatly overestimate the
extent of the bounce. As a result, we used the carpenter’s square to measure the
maximum extent of the bounce while several subjects with various weights (125 lbs, 177
lbs, and 203 lbs) ran on the treadmill at 6 stages of the Bruce protocol. The fluctuation
about the height of the front of the treadmill (H) was measured and converted to the
corresponding angle, and the results are displayed in Table 29. The maximum extent of
the bounce outside the MRI room was measured to be 0.176° while the maximum bounce
inside the MRI room was 0.146°; in both cases, this occurred with the 125 lb person at
stage 5. This is probably due to the fact that the 125 lb person, being shorter, needed to
run at stage 5 while the other two taller individuals were able to walk. These results show
that the treadmill is reasonably rigid, but further improvements can be made by
minimizing compliance in the system and improving the rigidity of the leg assembly,
such as by replacing aluminum with MR-compatible stainless steel.
Table 29: The extent of treadmill bounce during operation across a range of weights

<table>
<thead>
<tr>
<th>OUTSIDE MRI ROOM</th>
<th>125 Lbs</th>
<th>177 Lbs</th>
<th>203 Lbs</th>
</tr>
</thead>
<tbody>
<tr>
<td>H (in)</td>
<td>Bounce (±in)</td>
<td>Bounce (±deg)</td>
<td>Bounce (±in)</td>
</tr>
<tr>
<td>7.875</td>
<td>0.0313</td>
<td>0.0290</td>
<td>0.0938</td>
</tr>
<tr>
<td>9</td>
<td>0.0625</td>
<td>0.0582</td>
<td>0.0625</td>
</tr>
<tr>
<td>10.25</td>
<td>0.0625</td>
<td>0.0583</td>
<td>0.0938</td>
</tr>
<tr>
<td>11.375</td>
<td>0.0938</td>
<td>0.0877</td>
<td>0.1250</td>
</tr>
<tr>
<td>12.5625</td>
<td>0.1875</td>
<td>0.1761</td>
<td>0.1563</td>
</tr>
<tr>
<td>13.8125</td>
<td>0.1260</td>
<td>0.1178</td>
<td>0.1563</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>INSIDE MRI ROOM</th>
<th>125 Lbs</th>
<th>177 Lbs</th>
<th>203 Lbs</th>
</tr>
</thead>
<tbody>
<tr>
<td>H (in)</td>
<td>Bounce (±in)</td>
<td>Bounce (±deg)</td>
<td>Bounce (±in)</td>
</tr>
<tr>
<td>7.7813</td>
<td>0.0313</td>
<td>0.0290</td>
<td>0.0313</td>
</tr>
<tr>
<td>9</td>
<td>0.0313</td>
<td>0.0291</td>
<td>0.0625</td>
</tr>
<tr>
<td>10.2000</td>
<td>0.0625</td>
<td>0.0583</td>
<td>0.0625</td>
</tr>
<tr>
<td>11.375</td>
<td>0.0938</td>
<td>0.0877</td>
<td>0.0625</td>
</tr>
<tr>
<td>12.5000</td>
<td>0.1563</td>
<td>0.1467</td>
<td>0.0938</td>
</tr>
<tr>
<td>13.6875</td>
<td>0.0938</td>
<td>0.0883</td>
<td>0.0938</td>
</tr>
</tbody>
</table>

7.8.3 Noise Testing

Service software installed on the MRI system (Siemens Medical Solutions, Malvern, PA) was used to test for image interference caused by the treadmill system, and the combination of the treadmill and ECG systems. The treadmill system consisted of the treadmill itself as well as the hydraulic hoses and the fiber optic cables passing through a waveguide on the penetration panel in the MRI room. Additional components included a USB cable for Bluetooth communication with the keyboard and mouse inside the MRI room, and an ECG cable connected to the CAM 14 ECG module, both passing through a second waveguide leading to the MRI control room. The tests were performed using a
spherical phantom inside a larger cylindrical body phantom, with a power cord passed between the two phantoms to bring any noise into the magnet bore. The noise measurements were performed using the body coil. The tests included RF noise tests, RF spike checks, shim tests to check whether placing the treadmill immediately adjacent to the magnet affected the magnetic field homogeneity, and artifact tests to check for image artifacts in the sagittal, coronal, and transverse planes. The results are summarized in Table 30, where “−” indicated that the results were within the manufacturer’s specifications, and “+” indicated they were out of specification. The only test which was out of specification was the RF noise test with the entire setup connected, which exceeded specification in 10 out of 50 frequencies ranging from -250 kHz to 240 kHz about the center frequency in 10 kHz increments. The two potential sources of artifact are the shielded USB cable and the ECG cable passing through the waveguide, which can act as “antennas” to bring external RF noise into the MRI room, or can carry in noise directly from the PC or the ECG box located outside the MRI room. Since an independent test indicated that the shielded USB cable did not bring any noise into the MRI room, the likely source is the ECG cable. A potential solution for the noise from the PC or the ECG box is to create a switch to disconnect the ECG cable during imaging, while shielding of the ECG cable may minimize the “antenna” effect.
Table 30: Results of noise testing

<table>
<thead>
<tr>
<th></th>
<th>RF noise</th>
<th>RF spike check</th>
<th>Shim</th>
<th>Artifact</th>
</tr>
</thead>
<tbody>
<tr>
<td>Baseline</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Stationary treadmill + hoses + fiber optic cables</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Running treadmill + hoses + fiber optic cables</td>
<td>-</td>
<td>-</td>
<td>N/A</td>
<td>-</td>
</tr>
<tr>
<td>Stationary treadmill + hoses + fiber optic cables + ECG cable + USB cable</td>
<td>+</td>
<td>N/A</td>
<td>N/A</td>
<td>-</td>
</tr>
</tbody>
</table>

7.9 Bruce Protocol Testing in Human Subjects

Eight subjects completed the Bruce protocol test using the hydraulic treadmill positioned immediately adjacent to the MRI table. The first six subjects in Table 31 were healthy volunteers (average age 28 ± 8 years) and the last two were patients with known or suspected coronary artery disease (ages 19 and 42). In each case, the test was executed successfully. The belt started unassisted for all subjects ranging in weight between 110 lbs and 218 lbs, and both speed and elevation remained within the safety limits. The results, including the time to start imaging and the heart rate at the start of imaging expressed as beats per minute (bpm) and the percentage of maximum predicted heart rate based on age (%MPHR), are shown in Table 31. The healthy subjects exercised for an average of 13.3 ± 2.2 minutes, corresponding to stage 5 of the Bruce protocol, while both patients exercised for 9 minutes, corresponding to stage 3. As expected, the exercise time was shorter with patients than with healthy volunteers. Figure 103 shows the speed data acquired with subject #1, who reached peak stress at approximately the midpoint of stage
5. The corresponding elevation data for the same subject is shown in Figure 104. The upper and lower limits for speed and elevation are shown on the figures, illustrating the fact that both speed elevation remained within the limits. Figure 105 and Figure 106 show the speed and elevation data, respectively, for subject #7, who was a 218 lb cardiac patient. The imaging data obtained with the healthy subjects was analyzed and discussed in Section 4.2.

Table 31: Results of Bruce protocol testing in 6 healthy subjects and 2 cardiac patients

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Weight (lbs)</th>
<th>Total test time (min)</th>
<th>Bruce stage reached</th>
<th>Time to start cine (s)</th>
<th>Heart rate at rest (bpm; %MPHR)</th>
<th>Heart rate at peak exercise (bpm; %MPHR)</th>
<th>Heart rate at start of imaging (bpm; %MPHR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>23</td>
<td>110</td>
<td>13.7</td>
<td>5</td>
<td>34</td>
<td>64;32</td>
<td>187;95</td>
<td>152;77</td>
</tr>
<tr>
<td>2</td>
<td>21</td>
<td>125</td>
<td>12.0</td>
<td>4</td>
<td>23</td>
<td>72;36</td>
<td>176;88</td>
<td>134;67</td>
</tr>
<tr>
<td>3</td>
<td>18</td>
<td>125</td>
<td>12.7</td>
<td>5</td>
<td>23</td>
<td>70;35</td>
<td>196;97</td>
<td>157;78</td>
</tr>
<tr>
<td>4</td>
<td>34</td>
<td>165</td>
<td>10.0</td>
<td>4</td>
<td>24</td>
<td>66;35</td>
<td>193;104</td>
<td>164;88</td>
</tr>
<tr>
<td>5</td>
<td>36</td>
<td>160</td>
<td>15.0</td>
<td>5</td>
<td>21</td>
<td>54;29</td>
<td>181;98</td>
<td>177;96</td>
</tr>
<tr>
<td>6</td>
<td>33</td>
<td>165</td>
<td>16.3</td>
<td>6</td>
<td>24</td>
<td>60;32</td>
<td>173;93</td>
<td>161;86</td>
</tr>
<tr>
<td>7</td>
<td>19</td>
<td>218</td>
<td>9.0</td>
<td>3</td>
<td>28</td>
<td>70;35</td>
<td>173;86</td>
<td>120;60</td>
</tr>
<tr>
<td>8</td>
<td>42</td>
<td>200</td>
<td>9.0</td>
<td>3</td>
<td>36</td>
<td>53;63</td>
<td>155;87</td>
<td>112;63</td>
</tr>
</tbody>
</table>
Figure 103: Speed feedback obtained with a 110 lb healthy volunteer undergoing the Bruce protocol, with safety limits of ±0.6 mph indicated as dashed lines.

Figure 104: Elevation feedback obtained with a 110 lb healthy volunteer undergoing the Bruce protocol, with safety limits of ±1.13° indicated as dashed lines.
Figure 105: Speed feedback obtained with a 218 lb cardiac patient undergoing the Bruce protocol, with safety limits of ±0.6 mph indicated as dashed lines.

Figure 106: Elevation feedback obtained with a 218 lb cardiac patient undergoing the Bruce protocol, with safety limits of ±1.13° indicated as dashed lines.
In summary, this chapter presented the development and testing of a control system for a fully MRI-compatible water-hydraulic treadmill, with no ferromagnetic parts, electrical components, or electrical wires inside the MRI room. The speed and elevation sensing is performed directly on the treadmill using fiber optics, while the electronics and the PC with the LabVIEW control program are located outside the MRI room. The proportional-integral speed control is conducted using the electric motor controller, except for a special case at startup, while elevation control is performed through LabVIEW using pulse width modulation. FMEA analysis was conducted and safety features were implemented within the LabVIEW program during pretest, startup, and exercise. In addition, an emergency stop button and a speed limit for the electric motor were implemented. Speed and elevation performance testing was carried out with several subjects, and it was determined that the maximum speed error over all weights was 1.6% and the maximum elevation error over all weights was 2.5%. Finally, six healthy subjects and two cardiac patients successfully underwent the Bruce treadmill protocol, with speed and elevation remaining within the safety limits in each case. This treadmill is an essential element for making exercise CMR clinically practical, and enables us replicate the standard exercise echocardiography setup, but with the added imaging advantages of CMR.
CHAPTER 8

CONCLUSIONS AND FUTURE DIRECTIONS

This chapter will briefly summarize the work presented in this thesis and the main conclusions, followed by a discussion of future directions. The future work was divided into technical improvements, primarily pertaining to the MRI-compatible hydraulic treadmill, and new avenues for clinical research enabled by treadmill exercise CMR which were not previously available due to the limitations of existing stress imaging modalities.

8.1 Conclusions

This thesis presented the development and validation of a new cardiac stress imaging modality combining treadmill exercise stress testing with cardiac magnetic resonance imaging (exercise CMR).
The first step was to test the feasibility of exercise CMR, combining cardiac wall motion and myocardial perfusion obtained immediately post-exercise with myocardial viability at rest. The testing was performed in healthy subjects using a partially MRI-compatible treadmill located in the corner of the MRI room. Next, imaging optimization was carried out to compare the TSENSE and TGRAPPA methods of parallel imaging under the condition of deep breathing encountered after exercise stress, and both quantitative and qualitative results showed that TGRAPPA delivered superior artifact performance. TSENSE suffered from artifacts when the FOV was smaller than the object or the coil map was imperfect due to breathing motion. In addition, it was shown that using a 32-channel cardiac array coil enabled rate 4 parallel acceleration, resulting in improved temporal resolution.

ECG monitoring inside the MRI room was required both during treadmill exercise and recovery. Although the ECG was known to be non-diagnostic within the bore of any high-field magnet due to the magnetohydrodynamic (MHD) effect, the magnetic field threshold below which accurate ECG monitoring was feasible inside the MRI room but outside of the magnet bore was determined. It was shown that reliable ECG measurements could be obtained within the ST segment at magnetic field strengths below approximately 70 mT measured at the aortic arch, which corresponded to approximately 80 cm from the bore entrance for the Siemens 1.5T Avanto system, but could be extrapolated to any other system knowing the magnetic field plot. This indicated that accurate ECG monitoring was feasible during treadmill exercise inside the MRI room and recovery on the MRI patient table.
Next, the feasibility of exercise CMR for accurate diagnosis of ischemia was investigated in 43 patients with known or suspected coronary artery disease who were referred for treadmill nuclear imaging. Both exercise CMR and nuclear data were obtained while exercising the patient only once. It was shown that exercise CMR could accurately detect coronary artery disease compared to coronary angiography as the gold standard, and preliminary results indicated favorable accuracy compared to nuclear stress imaging.

Although the preliminary studies in healthy subjects and cardiac patients were conducted using a partially MRI-compatible treadmill in the corner of the MRI room, this would be impractical as a standard clinical stress testing modality. It was necessary to minimize the time between end of exercise and imaging in order to capture rapidly resolving exercise-induced cardiac wall motion abnormalities, necessitating treadmill placement beside the MRI table. Requiring the patient to walk from a treadmill positioned any distance from the MRI table would create a potential safety concern since patients may become lightheaded and subject to falling immediately following maximal exercise. Furthermore, there was risk of operator error in moving the treadmill too close to the magnet, and the entire setup in the corner of the room would not be possible at higher field strengths such as 3T. Therefore, a fully MRI-compatible hydraulic treadmill was developed, which could be positioned immediately adjacent to the MRI table. This thesis presented all aspects pertaining to the development and testing of the feedback control system for continuous control of speed and elevation. Component selection and design of speed and elevation control were presented, followed by LabVIEW software
design and implementation and the failure modes and effects analysis and implementation of safety features. Performance testing for speed and elevation was conducted over the range of speeds and elevations corresponding to 6 stages of the Bruce protocol. Finally, healthy subjects and cardiac patients exercised to peak stress on the hydraulic treadmill positioned immediately adjacent to the MRI table.

In summary, this thesis demonstrated that exercise CMR can be performed safely and accurately inside the MRI room. All necessary equipment was designed or selected to enable the stress testing team to remain inside the MRI room and in direct communication with the patient at all times, and to commence imaging as quickly as possible post-exercise. Although a comprehensive exercise CMR exam combining stress cardiac wall motion and stress myocardial perfusion with myocardial viability imaged at rest demonstrated favorable accuracy compared to nuclear stress imaging, the sample size would need to be increased in order to draw statistical conclusions. Successfully demonstrating that exercise CMR is diagnostically and prognostically superior to nuclear stress imaging in a larger clinical trial could have a significant impact on clinical practice and potentially change the current standard of cardiac stress testing.

In the subsequent sections, the future work in terms of potential technical improvements and new clinical applications will be discussed.
8.2 Technical Improvements

Some possible technical improvements with the hydraulic treadmill which could make it more feasible as a robust clinical tool, including digital elevation sensing or open-loop elevation control, and the integration of LabVIEW and Cardiosoft software, are presented in this section. In addition, some possible tools for dynamic speed and elevation performance validation are suggested.

Elevation Sensing and Control

Alternative methods of obtaining elevation feedback to consider include replacing the analog elevation sensor with digital sensing or open-loop control, as discussed below. The benefits of the analog sensor include its simplicity of operation, convenient and simple mounting arrangement, and the intrinsic ability to discern the direction of motion by providing a voltage output proportional to the treadmill elevation. However, it is associated with several drawbacks. In addition to the warm-up time required to reach the steady state output when power is turned on, another issue is its robustness. For example, if one of the mounts were to be accidentally moved due to impact, this could change the relationship between the sensor output and the treadmill angle, requiring realignment and recalibration. Although this is less of a concern in a research setting with trained staff who take significant care in handling the system and monitoring its performance, it becomes a greater concern in a fully clinical setting, especially at other institutions.
One alternative may be digital sensing using a strip of alternating bands and gaps as the target, which would be positioned between transmitting and receiving fibers mounted to the treadmill base and which would linearly translate with the cylinder stroke. The same photoelectric fiber optic switch as that used in speed and emergency stop detection could be used to count the pulses, starting at the lowest treadmill elevation as the “zero” position. However, a major challenge with this approach would be to discern the direction of motion, necessitating quadrature sensing. The implication of this is that two sensors would need to be used positioned 90° out of phase, each with a set of transmitting and receiving fibers placed as close as possible to the target in order to minimize the light beam expansion and maximize the resolution. This would make the mounting arrangement complex, as well as introduce another sensor and an additional set of fiber optic cables through the waveguide. Although the reliability may be somewhat improved, this approach would still be subject to misalignment error between the fibers and the target, between the transmitting and receiving fibers of each sensor, and between the two sets of fibers positioned 90° out of phase, and may increase the overall likelihood for component failure due to the additional components present in the system.

A second alternative may be an open-loop system, similar to the standard GE treadmills which use an electric motor coupled to a screw drive. This could potentially be accomplished through two flow meters which could regulate the volume of water released into the lift cylinder, and from the lift cylinder back into the reservoir. The flow meters could be placed outside of the MRI room, and would not need to be MRI-compatible. This type of approach should be feasible in a mechanically stiff system with
minimized compliance, such that the volume of water released into the lift cylinder is independent of subject weight. This method could potentially eliminate the need for the elevation sensor and the associated fiber optic cables or any other MRI-compatible components, greatly simplify component mounting and elevation control, and make the system significantly more robust in practice.

**LabVIEW/Cardiosoft Integration**

Another element which is required for exercise CMR to become fully clinically feasible is the integration of Cardiosoft software with the LabVIEW control program. Similar to the standard GE treadmills, the hydraulic treadmill should operate by using Cardiosoft to send the commands to start and stop the treadmill, as well as the target speed and elevation, to the LabVIEW program. After initially running the LabVIEW executable to start the program, the LabVIEW program should perform the treadmill control in the background with no clinician interaction, and should only become visible if the safety or warning limits have been exceeded.

Another aspect to consider is the remote desktop communication between the Cardiosoft PC and the LabVIEW PC, required because the LabVIEW PC needs to be located next to the DAQ in the equipment room and the Cardiosoft PC needs to be located in the MRI control room due to the restrictions in the length of the ECG cable. A way to enable a single PC located in the equipment room to be used for both may be to install a waveguide or a penetration panel near the floor of the MRI room, below the
current penetration panel. This would enable the ECG cable to reach the patient on the treadmill positioned immediately adjacent to the MRI table, and would allow Cardiosoft to operate on the LabVIEW PC. An additional benefit would be that this would shorten the required length of the hydraulic hoses and the fiber optic cables. However, if the treadmill is to be used at other institutions, the added cost and complexity of installing a new waveguide or penetration panel should be considered prior to shortening the hoses and cables.

**Dynamic Validation**

If external validation of dynamic performance of the speed control system is desired in addition to the steady state validation, an electrical 5th wheel device with continuous output which can be acquired by the PC could be used. The output should be acquired synchronously with the speed feedback obtained from the photoelectric fiber optic switch; ideally, the device would have a response time ≤0.25 seconds, matching the current speed feedback temporal resolution. In order to overcome the dynamic limitations of the electrolytic inclinometer used to validate the steady state elevation performance, an alternative such as a force balance accelerometer may be considered. As in dynamic speed sensing, the output would need to be acquired by the PC simultaneously with the elevation sensor output. However, the requirements for computer-readable output, fast response time, and relatively high accuracy drive up the cost of these components, which would only be used for dynamic validation and are not an integral part of the treadmill system.
8.3 New Clinical Applications

In addition to coronary artery disease detection, treadmill exercise CMR using the fully MRI-compatible treadmill may open new avenues for research in directions currently not available with the current stress testing modalities. The examples proposed here include right ventricle (RV) dysfunction, physiologic studies of the relationship between wall motion and perfusion, and investigation of the mechanisms of myocardial fibrosis in endurance athletes.

RV Dysfunction

The first example presented here is the application of exercise CMR for detecting right ventricle (RV) dysfunction by measuring the global RV function parameters (ejection fraction, cardiac output) at peak exercise stress. The complex RV shape and orientation combined with poor endocardial definition severely complicate the assessment of RV function using echocardiography. In contrast, CMR is a 3D technique which can acquire images of virtually any plane in the body, and has been established as the gold standard for quantitative assessment of RV volumes, ejection fraction, and mass [110-112]. Although it has previously been shown that multi-detector row computed tomography (CT) is comparable to CMR in quantifying RV volumes and ejection fraction in patients with Tetralogy of Fallot or Transposition of the Great Arteries [113], CT is limited by radiation exposure and insufficient temporal resolution to image the rapid heart rates associated with exercise stress.
Exercise CMR could be performed in patients referred for cardiopulmonary testing due to known or suspected chronic RV pressure overload resulting from pulmonary hypertension, heart transplant, Tetralogy of Fallot, Transposition of the Great Arteries, or other disease states. Cardiopulmonary testing determines changes in oxygen consumption (and indirectly metabolism) during exercise by measuring the oxygen and carbon dioxide content in expired air. The peak oxygen consumption (VO\textsubscript{2}max) is the point when the oxygen consumption curve reaches a plateau despite progressive increments in workload, and is a standard indicator of disease state. The global RV function parameters may be correlated with VO\textsubscript{2}max to demonstrate that exercise CMR is able to detect RV dysfunction.

An important specific potential application is the optimal timing of pulmonary valve replacement after Tetralogy of Fallot repair. The tradeoffs are the intrinsic morbidity of the procedure and the need for repeat valve replacement in the younger patient cohort versus ensuring that RV dysfunction has not progressed beyond the “point of no return”. The criteria for pulmonary valve replacement include both RV ejection fraction as measured by CMR and exercise intolerance, currently requiring two separate exams (exercise stress test and CMR) to be performed.

In summary, the combination of exercise testing and CMR may be the ideal modality to safely and non-invasively image the RV response to exercise, which is largely unknown due to the technical limitations of existing cardiac imaging techniques.
This new modality may therefore shed new light on the physiology of right heart pressure overload caused by a variety of pathological conditions.

**Physiologic Studies of Wall Motion and Perfusion**

Exercise CMR may open avenues for physiologic studies such as the comparison of wall motion and perfusion for ischemia detection immediately after exercise. Factors such as the overall relationship and correlation, the time course, and the relative extent of the wall motion and perfusion defects would be highly valuable. For example, the severity/extent of a perfusion defect needed before there is a wall motion abnormality is an important physiologic question that could be answered by exercise CMR and cannot be addressed any other way. These factors would have broad implications for stress testing in general.

**Myocardial Fibrosis in Endurance Athletes**

Endurance athletes have 3x higher risk of sudden cardiac death, mostly occurring during physical exercise [114]. Conduction abnormalities are so frequent in these athletes that they are considered “normal”, but can lead to adverse cardiac events long-term. Autopsy findings in athletes reveal myocardial fibrosis, which increases the likelihood to develop arrhythmias and conduction blocks. Furthermore, troponin levels are known to be elevated during competitive events.
Although the underlying cause of myocardial fibrosis in athletes is unclear, the elevated troponin levels suggest the presence of ischemia. Treadmill exercise CMR may provide an excellent method to investigate the mechanisms of fibrosis. Elite athletes may be asked to undergo an extended treadmill test to simulate an endurance event, accompanied by continuous 12-lead ECG to monitor ischemic changes, immediately followed by stress perfusion imaging to determine if there are regions of perfusion defects. This could be followed by delayed enhancement imaging to check for and quantify the presence of fibrosis; CMR is the ideal tool for this since echocardiography does not provide viability information and nuclear imaging has inadequate spatial resolution to capture the limited extent of fibrosis expected in these athletes. Delayed enhancement imaging may be followed by edema imaging and quantification in intervals to investigate the presence and progression of myocyte damage. Following the CMR imaging, a blood sample may be drawn to check troponin levels. Combining all this information obtained from a treadmill CMR test may provide significant insights regarding the mechanisms underlying myocardial fibrosis in athletes.
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M. S. Cocker, Strohm, O., Smith, D.J., Butler, C., Belenkie, I., Meeuwisse, W., Friedrich, M. G., "Elite olympic calibre high-endurance athletes have evidence for myocardial fibrosis: a cardiovascular magnetic resonance study," in International Society for Magnetic Resonance in Medicine, Hawaii, 2009.
APPENDIX A: Motor Controller Parameters

Table 32: Summary of the motor controller parameters which have been modified and/or are relevant to proportional-integral speed control

<table>
<thead>
<tr>
<th>#</th>
<th>Name &amp; Description</th>
<th>Default</th>
<th>Current</th>
</tr>
</thead>
<tbody>
<tr>
<td>41</td>
<td>Motor NP Volts</td>
<td>N/A</td>
<td>480 V</td>
</tr>
<tr>
<td>42</td>
<td>Motor NP FLA</td>
<td>N/A</td>
<td>6.4 A</td>
</tr>
<tr>
<td>43</td>
<td>Motor NP Hz</td>
<td>N/A</td>
<td>60 Hz</td>
</tr>
<tr>
<td>44</td>
<td>Motor NP RPM</td>
<td>N/A</td>
<td>1700 RPM</td>
</tr>
<tr>
<td>45</td>
<td>Motor NP Power</td>
<td>N/A</td>
<td>4.7 HP</td>
</tr>
<tr>
<td>53</td>
<td>Motor Ctrl Select (selects type of electric motor control)</td>
<td>0 (sensorless vector)</td>
<td>2 (custom V/Hz)</td>
</tr>
<tr>
<td>55</td>
<td>Maximum Freq (that the drive will output)</td>
<td>110 or 130 Hz</td>
<td>100 Hz</td>
</tr>
<tr>
<td>80</td>
<td>Feedback Select (for electric motor)</td>
<td>0 (open loop)</td>
<td>0 (open loop)</td>
</tr>
<tr>
<td>82</td>
<td>Maximum Speed (speed reference high)</td>
<td>50 or 60 Hz</td>
<td>100 Hz</td>
</tr>
<tr>
<td>83</td>
<td>Overspeed Limit (Max Speed + Overspeed Lmt ≤ Max Freq)</td>
<td>10 Hz</td>
<td>0 Hz</td>
</tr>
<tr>
<td>90</td>
<td>Speed Ref A Sel Controls source of desired speed for electric motor</td>
<td>2 (Analog In 2)</td>
<td>1 (Analog In 1)</td>
</tr>
<tr>
<td>91</td>
<td>Speed Ref A High Upper limit of speed reference</td>
<td>50 or 60 Hz</td>
<td>100 Hz</td>
</tr>
<tr>
<td>124</td>
<td>PI Configuration</td>
<td>Excl Mode = 0</td>
<td>Excl Mode = 1</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Preload = 0</td>
<td>Preload = 1</td>
</tr>
<tr>
<td>125</td>
<td>PI Control</td>
<td>PI Enable = 0</td>
<td>PI Enable = 1</td>
</tr>
<tr>
<td>126</td>
<td>PI Reference Select</td>
<td>0 (PI Setpoint)</td>
<td>1 (Analog In 1)</td>
</tr>
<tr>
<td>128</td>
<td>PI Feedback Select</td>
<td>2 (Analog In 2)</td>
<td>8 (Encoder)</td>
</tr>
<tr>
<td>129</td>
<td>PI Integral Time</td>
<td>2.00 s</td>
<td>2.00 s</td>
</tr>
<tr>
<td>130</td>
<td>PI Proportional Gain</td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>131</td>
<td>PI Lower Limit</td>
<td>-100% (of Max Freq)</td>
<td>-100% (of Max Freq)</td>
</tr>
</tbody>
</table>

Continued
Table 32 Continued

<table>
<thead>
<tr>
<th>#</th>
<th>Name &amp; Description</th>
<th>Default</th>
<th>Current</th>
</tr>
</thead>
<tbody>
<tr>
<td>132</td>
<td>PI Upper Limit</td>
<td>100% (of Max Freq)</td>
<td>100% (of Max Freq)</td>
</tr>
<tr>
<td>133</td>
<td>PI Preload</td>
<td>100%</td>
<td>100%</td>
</tr>
<tr>
<td>140</td>
<td>Accel Time 1</td>
<td>10.0 s</td>
<td>3.0 s</td>
</tr>
<tr>
<td>142</td>
<td>Decel Time 1</td>
<td>10.0 s</td>
<td>3.0 s</td>
</tr>
<tr>
<td>190</td>
<td>Direction Mode</td>
<td>0 (unipolar)</td>
<td>2 (reverse disable)</td>
</tr>
<tr>
<td>322</td>
<td>Analog In 1High</td>
<td>10 V</td>
<td>5 V</td>
</tr>
<tr>
<td>323</td>
<td>Analog In 1 Low</td>
<td>0 V</td>
<td>0 V</td>
</tr>
<tr>
<td>361</td>
<td>Digital Input 1</td>
<td>4 (Stop-CF)</td>
<td>0</td>
</tr>
<tr>
<td>362</td>
<td>Digital Input 2</td>
<td>5 (Start)</td>
<td>7 (Run)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Starts &amp; stops motor</td>
</tr>
<tr>
<td>363</td>
<td>Digital Input 3</td>
<td>18 (Auto/Manual)</td>
<td>26 (PI Enable)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Turns PI control on &amp; off</td>
</tr>
<tr>
<td>364</td>
<td>Digital Input 4</td>
<td>15 (Speed Sel 1)</td>
<td>1 (Enable)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Emergency stop</td>
</tr>
<tr>
<td>365</td>
<td>Digital Input 5</td>
<td>16 (Speed Sel 2)</td>
<td>1 (Enable)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Electric motor speed safety</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>shutoff</td>
</tr>
<tr>
<td>366</td>
<td>Digital Input 6</td>
<td>17 (Speed Sel 3)</td>
<td>1 (Enable)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Water level switch</td>
</tr>
<tr>
<td>380</td>
<td>Digital Output 1</td>
<td>1 (Fault)</td>
<td>10 (At Freq)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Changes state to OFF when freq in</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>param 381 is reached</td>
</tr>
<tr>
<td>381</td>
<td>Digital Output 1 Level</td>
<td>0</td>
<td>70 Hz</td>
</tr>
<tr>
<td>412</td>
<td>Motor Feedback Type</td>
<td>0 (quadrature)</td>
<td>2 (single check)</td>
</tr>
<tr>
<td>413</td>
<td>Encoder PPR</td>
<td>1024 PPR</td>
<td>4 PPR</td>
</tr>
<tr>
<td>460</td>
<td>PI Reference High</td>
<td>100 %</td>
<td>100 %</td>
</tr>
<tr>
<td>461</td>
<td>PI Reference Low</td>
<td>-100 %</td>
<td>0</td>
</tr>
<tr>
<td>462</td>
<td>PI Feedback High</td>
<td>100 %</td>
<td>100 %</td>
</tr>
<tr>
<td>463</td>
<td>PI Feedback Low</td>
<td>0</td>
<td>-100 %</td>
</tr>
</tbody>
</table>
APPENDIX B: Failure Mode Testform

Exercise MRI – Hydraulic Treadmill Control System Testform

Test performed by:  
Date:  
Signature:

<table>
<thead>
<tr>
<th>Test</th>
<th>Expected Outcome</th>
<th>Comments</th>
</tr>
</thead>
</table>
| P1 1. Disconnect one of fiber optic quick disconnects leading to elevation sensor  
2. Start pretest                                                   | _Treadmill remains at current elevation  
_ Motor is stopped  
_ Dialog box pops up informing the user of failure ("No elevation feedback")  
_ After "OK" is pressed on the dialog box, a 2nd dialog box pops up ("Elevation below limits at end of pretest. Try restarting pretest.") |                                                                                                                                              |
| P1 1. Use a piece of paper to obstruct light path between transmitting and receiving fibers of elevation sensor  
2. Start pretest                                                      | _Treadmill remains at current elevation  
_ Motor is stopped  
_ Dialog box pops up informing the user of failure ("No elevation feedback")  
_ After "OK" is pressed on the dialog box, a 2nd dialog box pops up ("Elevation below limits at end of pretest. Try restarting pretest.") |                                                                                                                                              |
| P1 1. Disconnect cable labeled “ELEVATION SENSOR POWER” from fuse box  
2. Start pretest                                                      | _Treadmill remains at current elevation  
_ Motor is stopped  
_ Dialog box pops up informing the user of failure ("No elevation feedback")  
_ After OK is pressed on the dialog box, a 2nd dialog box pops up ("Elevation below limits at end of pretest. Try restarting pretest.") |                                                                                                                                              |
| P2 1. Unplug power cable to raise valve from solenoid box  
2. Start pretest                                                      | Once the treadmill raises 1' above desired pretest elevation (see pretest panel)  
_ Motor stops  
_ Treadmill stops and lowers slowly to zero elevation  
_ Dialog box pops up informing the user of failure ("Elevation above limits") |                                                                                                                                              |
| P3 1. Unplug power cable to lower valve from solenoid box  
2. Start pretest                                                      | _Treadmill does not elevate to target  
_ After end of pretest, dialog box pops up warning the user ("Elevation below limits at end of pretest. Try restarting pretest.") |                                                                                                                                              |
| P3 | 1. Start pretest  
2. Stop pretest prematurely (within 1 sec) | ___Treadmill does not elevate  
___Dialog box pops up warning the user (“Elevation below limits at end of pretest. Try restarting pretest.”) |
| P4 | 1. Unplug power cable to accumulator valve from solenoid box  
2. Start pretest | ___After 2 pulses detected (belt will move ~5 sec), motor stops  
___Treadmill remains at current elevation  
___Dialog box pops up (“Speed feedback detected during pretest. Check accumulator valve”)  
___After OK is pressed on the dialog box, a 2nd dialog box pops up (“Elevation below limits at end of pretest”) |

### STARTUP

<table>
<thead>
<tr>
<th>Test</th>
<th>Expected Outcome</th>
<th>Comments</th>
</tr>
</thead>
</table>
| **S1.1** | 1. Disconnect one of fiber optic quick disconnects leading to speed switch  
2. Start test | ___Belt moves at low speed (≤ stage 1)  
___Program times out and motor stops  
___Dialog box pops up informing the user of timeout due to zero speed feedback | |
| **S1.2** | 1. Disconnect cable labeled “SPEED SENSOR POWER” from fuse box  
2. Start test | ___Belt moves at low speed (≤ stage 1)  
___Program times out in 30 sec and motor stops  
___Dialog box pops up informing the user of timeout due to zero speed feedback | |
| **S2.1** | 1. Disconnect hose leading to hydraulic motor labeled “FWD”  
2. Start test | ___Belt remains stationary  
___Program times out in 30 sec and motor stops  
___Dialog box pops up informing the user of timeout due to zero speed feedback | |
| **S2.2** | 1. Disconnect drive shaft coupler  
2. Start test | ___Belt remains stationary or moves slightly due to slipping  
___Program times out in 30 sec and motor stops, or test starts and speed feedback below limits is detected  
___Dialog box pops up informing the user of timeout due to zero speed feedback, or dialog box pops up “Speed feedback below limits” | |
| **S2.3** | 1. Disconnect return hose labeled “RETURN”  
2. Start test | ___Belt remains stationary or moves very slowly  
___Program times out in 30 sec and motor stops, or test starts and speed feedback below limits is detected  
___Dialog box pops up informing the user of timeout due to zero speed feedback, or dialog box pops up “Speed feedback below limits” | |

### EXERCISE

<table>
<thead>
<tr>
<th>Test</th>
<th>Expected Outcome</th>
<th>Comments</th>
</tr>
</thead>
</table>
| **T1.1** | 1. Start test  
2. At any point, disconnect one of fiber optic quick disconnects leading to speed switch | ___Motor controller stops  
___Treadmill stops  
___Dialog box pops up informing the user of failure (“Speed feedback below limits”) | |
| **T1.2** | 1. Start test  
2. At any point, disconnect cable labeled | ___Motor controller stops  
Treadmill stops |
<table>
<thead>
<tr>
<th>Test</th>
<th>Description</th>
<th>Expected Outcome</th>
</tr>
</thead>
<tbody>
<tr>
<td>T3.1</td>
<td>Start test, disconnect one fiber optic quick disconnects to elevation sensor</td>
<td>Motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T3.2</td>
<td>Start test, use a piece of paper to obstruct light path between fibers of elevation sensor</td>
<td>Motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T3.3</td>
<td>Start test, disconnect cable labeled &quot;ELEVATION SENSOR POWER&quot; from fuse box</td>
<td>Motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T4</td>
<td>Start test, unplug cable to raise valve from solenoid box</td>
<td>Motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T6</td>
<td>Start test, unscrew reservoir cap and drain water level</td>
<td>Water level indicator starts blinking, motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T7</td>
<td>Start test, unscrew reservoir cap and drain water level</td>
<td>Water level indicator starts blinking, motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T8</td>
<td>Start test, unscrew reservoir cap and drain water level</td>
<td>Water level indicator starts blinking, motor controller stops, treadmill stops, dialog box informs failure.</td>
</tr>
<tr>
<td>T11</td>
<td>Start test, unplug cable to lower valve from solenoid box</td>
<td>Treadmill lowers fully, elevation feedback indicator starts blinking, motor stops, treadmill stops.</td>
</tr>
<tr>
<td>T12</td>
<td>Turn off safety feature “Stop motor if elevation below limits”</td>
<td>Treadmill lowers, after 5 sec, elevation feedback indicator starts blinking.</td>
</tr>
<tr>
<td></td>
<td>Start test, unplug cable to lower valve from solenoid box</td>
<td>After 5 sec, elevation feedback indicator starts blinking.</td>
</tr>
<tr>
<td></td>
<td>Start test, enter manual mode at 11°</td>
<td>After 5 sec, elevation feedback indicator starts blinking.</td>
</tr>
<tr>
<td></td>
<td>Enter manual mode at 11°</td>
<td>After 5 sec, elevation feedback indicator starts blinking.</td>
</tr>
</tbody>
</table>
APPENDIX C: LabVIEW Code and User Interface

Figure 107: Detailed description of the function of each button on the LabVIEW user interface within the Cardiosoft screen
Figure 108: Loop timing
Figure 109: Automatic execution of Bruce protocol by assigning the speed and elevation values based on the time elapsed since the start of the test. The top shows the situation of the normal Bruce protocol, and the bottom shows the case where the manual mode button is pressed; in the latter case, the commanded Bruce protocol values are replaced by those manually entered by the user.
Figure 110: Conversion from pulse count to belt speed

Figure 111: Raise valve error within pulsed error limits (<0.0825°); no control applied

Figure 112: Raise valve error exceeds both pulsed and continuous error limits (>0.935°); continuous control applied
Figure 113: Raise valve exceeds pulsed error limits but is within continuous error limits;

pulsed control applied
Figure 114: Computation of speed safety limits, checks for whether the speed feedback is within the limits, and the error messages displayed if the feedback is above or below the limits
Figure 115: Stopping the motor controller if the elevation feedback is either above or below the limits
APPENDIX D: Raw Data from Speed Performance Testing

Outside MRI Room

<table>
<thead>
<tr>
<th>Stage</th>
<th>Unloaded</th>
<th>125 Lbs</th>
<th>177 Lbs</th>
<th>203 Lbs</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Pulse Tachometer (no. pulses)</td>
<td>5th Wheel (ft)</td>
<td>Pulse Tachometer (no. pulses)</td>
<td>5th Wheel (ft)</td>
</tr>
<tr>
<td>1</td>
<td>64</td>
<td>74.6</td>
<td>64</td>
<td>75</td>
</tr>
<tr>
<td>2</td>
<td>92</td>
<td>109.6</td>
<td>93</td>
<td>109.5</td>
</tr>
<tr>
<td>3</td>
<td>127</td>
<td>148.1</td>
<td>126</td>
<td>148.5</td>
</tr>
<tr>
<td>4</td>
<td>155</td>
<td>183.9</td>
<td>156</td>
<td>182.5</td>
</tr>
<tr>
<td>5</td>
<td>187</td>
<td>220</td>
<td>185</td>
<td>218</td>
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<tr>
<td>6</td>
<td>205</td>
<td>243.5</td>
<td>204</td>
<td>240.8</td>
</tr>
</tbody>
</table>

Inside MRI Room

<table>
<thead>
<tr>
<th>Stage</th>
<th>Unloaded</th>
<th>125 Lbs</th>
<th>177 Lbs</th>
<th>203 Lbs</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Pulse Tachometer (no. pulses)</td>
<td>5th Wheel (ft)</td>
<td>Pulse Tachometer (no. pulses)</td>
<td>Pulse Tachometer (no. pulses)</td>
</tr>
<tr>
<td>1</td>
<td>64</td>
<td>74.4</td>
<td>63</td>
<td>64</td>
</tr>
<tr>
<td>2</td>
<td>92</td>
<td>109.2</td>
<td>92</td>
<td>92</td>
</tr>
<tr>
<td>3</td>
<td>127</td>
<td>149.1</td>
<td>127</td>
<td>126</td>
</tr>
<tr>
<td>4</td>
<td>154</td>
<td>185.7</td>
<td>154</td>
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<tr>
<td>5</td>
<td>186</td>
<td>217.3</td>
<td>184</td>
<td>185</td>
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<tr>
<td>6</td>
<td>202</td>
<td>240</td>
<td>203</td>
<td>203</td>
</tr>
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