INFRARED FRACTURE DETECTION OF BARE METAL STENTS

by

Austen Kolar Scudder

A Thesis Submitted in
Partial Fulfillment of the
Requirements for the Degree of
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in Engineering
at
The University of Wisconsin-Milwaukee
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ABSTRACT

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The University of Wisconsin-Milwaukee, 2011
Under the Supervision of Professor Ilya V. Avdeev

Bare metal stents are commonly used as an alternative to Coronary Bypass Surgery (CABG), however they are susceptible to fracture, leaving the patient prone to restenosis. Traditional stents are designed to be flexible, yet strong enough to compress the surrounding plaque and support the local vasculature. Optimization of stent designs has become a substantial goal as they are deployed in ever-increasing dynamic conditions, particularly peripheral arteries. Current testing methods involve accelerated fatigue testing, which involves millions of cycles in various deformation modes. Few, if any, checkpoints occur until the test is complete, leaving engineers unaware of the true magnitude of its robustness. A novel method is being presented which will demonstrates the potential for non-invasive feedback on a stent's mechanical condition. The method utilizes electrical theory by detecting change in resistance in a structure as well as observing infrared light emissions from the electrically heated stent device. Initially numerical models were generated to simulate the electrical and thermal responses. These models were validated by experimental results, showing similar infrared light emission patterns (temperature maps). As a result, the approach provides the ability to identify the location of the respective fracture. This allows the test to continue unaffected, providing greater insight into fracture sequences. Understanding precise fracture chronology and
topology offers the ability to associate failures with previously developed mechanical simulations. Furthermore, areas of accumulated fatigue stress can be identified and validated, and engineers can optimize features in subsequent design iterations.
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Chapter 1: Introduction

1.1 Motivation and Objectives

Atherosclerosis is the most prevalent form or arterial disease, (Noble et al. 2005) and accounts for approximately half of the deaths in the western world [1]. It is the result of excessive plaque accumulation, a product of calcium, cholesterol, fats and other substances found in blood. This accumulation impedes flow volume through the vascular tissue and can lead to serious health complications, including heart attack, stroke and potentially death. Similarly, arteriosclerosis, occurs when the artery walls become hardened and/or narrowed. This can be a direct result of atherosclerosis and result in the same life threatening conditions.

Previous methods for treating these arterial diseases have depended on the specific location and severity of the problem. When the severity is not excessive, physicians may be able to prescribe medications. However, this is not always a sustainable option. Beginning in the 1960’s, surgeons began to perform coronary arterial bypass surgery (CABG), which posed a new set of risks for patients. A bypass surgery is a highly invasive and delicate procedure. Traditionally, the heart is stopped during this procedure and involves the use of machines to maintain blood circulation and oxygen flow through the body, known as "on-pump." A newer technique, referred to as "off-pump" CABG, has been developed which is less invasive however standard complications remain, including risk of infection, surgical error and the need for general anesthesia. Neither approach yields significant benefits over the other [2].
The various degrees of atherosclerosis can be seen above. Mild atherosclerosis is inherent in an aging population, however severe atherosclerosis is becoming more prevalent. Another method known as angioplasty is a less invasive alternative to the CABG procedure. It involves inserting a catheter into the artery and inflating a balloon to compress the plaque material and widen the channel for blood flow. However the results are rarely as successful in the long term and require numerous reinterventions [3]. Moreover, this approach requires substantially greater reliance on medication. Though these methods may be better suited for different patients, the relative prognosis and quality of life is not significantly different [4].

A newer development in treating atherosclerosis incorporates stents, a cage-like structure percutaneously deployed and typically in conjunction with an angioplasty procedure. The stent serves to compress the plaque material to the vascular wall and maintain an open channel for blood flow. Stents can be deployed in numerous vascular locations, including the coronary arteries as well as the peripheral arteries. It has become
an efficient method for treating both Coronary Artery Disease (CAD) and Peripheral Artery Disease (PAD) due in large part to its minimally invasive approach.

Peripheral artery stenting has become the safest and most prevalent clinical solution for patients with atherosclerosis. However, several problems are commonly associated with stenting, particularly in relation to deployment in peripheral arteries such as the carotid and femoral arteries. Because peripheral arteries inherently experience tortuous deformations, extreme mechanical forces are imparted on the stent. Subsequently, stresses experienced in these stents occur at great magnitudes and high frequencies resulting in a substantial rate of failure. Furthermore, stent failure can be a catalyst for scar tissue growth and eventual restenosis.

In the past decade, stents have become physician's preferred approach for treating atherosclerosis. Though coronary stents have been widely successful, peripheral artery stents have posed a significant challenge to physicians. Because of the numerous types of stent geometries as well as the variant nature of patient vasculature, it can be difficult to provide the optimal stenting solution in every situation. What's more, an idealized scenario can still pose limitations based upon the stent's natural fatigue limit.

As common as stents are, the critical design factors are poorly understood. Because of their complex nature, and intricate structure, experimental testing methods can never fully encompass the potential loading conditions it will experience. To account for this, experimental testing typically uses an artificially high cycle count during validation. Unfortunately this process is slow and resource intensive. Moreover, the information is incomplete and generally unhelpful for future design iterations because of its "pass or fail" metric.
Additionally, because of the extensive detail in the structure, it also not easily described by analytical models. Finite Element Analysis (FEA) offers great potential to describe localized stresses in stent designs, and can achieve relatively quick results throughout a plethora of loading conditions. Moreover, these loading conditions have the potential to be customized on a patient-by-patient basis in order to assess what design and/or geometry is ideal for a given subject. Yet, substantial research remains to ascertain the correlation between FEA simulation and experimental fatigue tests.

An important factor in determining this correlation is understanding when and where a fracture occurs during experimental testing. This has been a point of interest for physicians, engineers and researchers alike. Previously developed In Vivo methods include Angiography, Intra Vascular Ultrasound (IVUS), and Computed Tomography (CT). However, these methods are unsuitable for characterizing specific design weaknesses or stress points as it is impossible to accurately quantify the range of motion and frequency of which a patient has taxed a given stent. Additionally, patient check-ups occur far too infrequently to provide adequate time or cycle resolution on a stent fracture or failure.

Ong and Serruys announced in a keynote address in 2005 that over 2.5 million stents are implanted on patients in [the] USA per year [5]. Due to the wide range of physiological conditions, stent design and development have correspondingly become more complex and rigorous. For example, a stented femoropopliteal Arterial Segment under 90/90 knee/hip flexion undergoes up to 69° bending, 25% of compression strain, while the proximal popliteal artery experiences up to a maximum of 29° bending and 9%
of compression strain [6,7]. The Internal Carotid Artery experiences up to 69.48° of torsion and 52.2° of bending [8].

Stent designs, almost exclusively generated in 3D CAD software, are translated into finite element analysis programs for a large variety of test simulations. Finite element analysis has been used in the biomechanical development since 1972 (Huiskes and Chao 1983), yet biological structures and desired clinical applications are difficult and require a thorough understanding of the problem and related complexities [9].

Use of finite element methods in the development of stents leads to a reduction in design iterations and improvements in quality and performance [10,11]. Yet no approach is fully capable of replacing physical testing, nor universally describing vascular interactions. However, significant research efforts have been devoted to addressing these issues, and many promising approaches have been devised.

The latest developments in imaging technology provide vascular geometry on an individual basis, creating the possibility of patient customizable devices. Eiho et al. presented a shape measurement method of aneurismal aorta for assisting CAD design of stents [12]. 2-D fluoroscopic images do not provide 3-D spatial information. To overcome the shortcomings of 2-D image data obtained from a fluoroscopic image of the aorta, the authors developed a method of registering a 2-D fluoroscopic image to a preoperative CT 3-D image. This 3-D/2-D registration is called digitally reconstructed radiography (DRR). This method is based on physical relationship between voxel value in a CT image and pixel value in fluoroscopic image. Preoperative CT data was registered with an interoperative fluoroscopic image by pattern matching. The M-estimator method was used. Many other methods have been developed to generate patient
specific FE models [13,14]. Alternatively, interacting devices can be modeled to optimize their dynamics. A CYPHER stent was modeled with two different balloon folding techniques (3 fold and 6 fold) in ABAQUS. Results showed that expansion behavior was sensitive to balloon folding technique and positioning [15,16,17]. Vascular injury can occur from high inflation pressures resulting in contact stresses between stent struts and vascular tissue. Improving the understanding of stent geometry and balloon inflation will serve to improve stent deployment and prevent or mitigate vascular injuries [18]. In the following section we will review underlying factors and considerations for stent design, including vascular dynamics, testing modalities and deployment procedures.

1.2 Literature Review

1.2.1 Review of Clinical Data

In order to accurately investigate the fatigue strength of any particular nitinol stent, an assessment of the dynamic range of motion within the human vasculature must be evaluated. Through the analysis of vascular strain, the equivalent mode of stress can be equated and characterized. Because stents vary in size and length, the strain has been characterized in a unitless system when possible. It is also important to recognize that unstented arteries are more flexible and yielding than stented arteries. From this it can be said that the maximum dynamic range of the stent will always be less than the range of an unstented artery, thus setting a limit for fatigue testing. However, it should also be noted that this can result in severe artery angulation at the distal and proximal stent junctions.

A collection of research papers were analyzed to generalize dynamic characteristics of femoral and carotid arteries. Of particular interest was the maximum
deflection during full range of motion (ROM). This data has been consolidated (see table) and can be used to define parameters for stent fatigue testing.

Within the superficial femoral artery there are several different regions of concern that exhibit different characteristics during hip and knee flexion. These include the distal SFA (proximal popliteal artery); middle SFA; and the distal popliteal artery. Though the primary modes of deformation appeared to be axial and bending, one study had cited a significant value for angular twist during hip and knee flexion. The majority of the axial deformations were negative, however positive strain (result of tension) was demonstrated in some cases as a result of knee/hip flexion.
<table>
<thead>
<tr>
<th>Unstented Arteries</th>
<th>Stented Arteries</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distal SFA</td>
<td>Middle SFA</td>
</tr>
<tr>
<td>Proximal SFA</td>
<td>Popliteal Artery</td>
</tr>
<tr>
<td>Popliteal Artery</td>
<td>Distal SFA</td>
</tr>
<tr>
<td>(90/90 Knee/hip</td>
<td>Proximal SFA</td>
</tr>
<tr>
<td>Flexion)</td>
<td>Popliteal Artery</td>
</tr>
<tr>
<td>Distal SFA</td>
<td>Middle SFA</td>
</tr>
<tr>
<td>Popliteal Artery</td>
<td>Popliteal Artery</td>
</tr>
<tr>
<td>(90/90 Knee/hip</td>
<td>(90/90 Knee/hip</td>
</tr>
<tr>
<td>Flexion)</td>
<td>Flexion)</td>
</tr>
</tbody>
</table>

- **Compression (70/20)**
  - Unstented: 14 ± 5%
  - Stented: 3 ± 2%

- **Compression (90/90)**
  - Unstented: 5.9 ± 3.0%
  - Stented: 2.8 ± 2.1%

- **Compression (90/90)**
  - Unstented: 23 ± 2%
  - Stented: 11 ± 5%

- **Compression (90/90)**
  - Unstented: 13 ± 11%
  - Stented: 10.2 ± 4.6%

- **Artery Max**
  - Unstented: 25%
  - Stented: 16%

- **Mode Max**
  - Unstented: 25%
  - Stented: 16%

- **Overall Max**
  - Unstented: 25%
  - Stented: 16%

*Table 1.1. Range of compressive deformation for popliteal arteries.*
<table>
<thead>
<tr>
<th>Artery Location</th>
<th>Unstented Arteries</th>
<th>Stented Arteries</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Distal SFA</td>
<td>Middle SFA</td>
</tr>
<tr>
<td></td>
<td>Proximal SFA</td>
<td>Popliteal Artery (90/90 Knee/Hip Flexion)</td>
</tr>
<tr>
<td>Bending (70/20)</td>
<td>11 ± 12 deg</td>
<td>3 ± 4 deg</td>
</tr>
<tr>
<td>Bending (90/90)</td>
<td>15 ± 14 deg</td>
<td>4 ± 3 deg</td>
</tr>
<tr>
<td>Bending (Angle)</td>
<td>8 ± 4 deg</td>
<td>5 ± 2 deg</td>
</tr>
<tr>
<td>Artery Max</td>
<td>NA</td>
<td>63 deg</td>
</tr>
<tr>
<td>Mode Max</td>
<td>63 deg</td>
<td>80 deg</td>
</tr>
<tr>
<td>Overall Max</td>
<td>80 deg</td>
<td></td>
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Table 1-2. Range of bending (angle measured) deformation for popliteal arteries.

<table>
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<th>Artery Location</th>
<th>Unstented Arteries</th>
<th>Stented Arteries</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Distal SFA</td>
<td>Middle SFA</td>
</tr>
<tr>
<td></td>
<td>Proximal SFA</td>
<td>Popliteal Artery (90/90 Knee/Hip Flexion)</td>
</tr>
<tr>
<td>Bending (Radius)</td>
<td>55 ± 10 mm</td>
<td>138 ± 53 mm</td>
</tr>
<tr>
<td>Bending (Radius)</td>
<td>13 mm</td>
<td></td>
</tr>
<tr>
<td>Artery Max</td>
<td>NA</td>
<td>13 mm</td>
</tr>
<tr>
<td>Mode Max</td>
<td>13 mm</td>
<td>191 mm</td>
</tr>
<tr>
<td>Overall Max</td>
<td>191 mm</td>
<td></td>
</tr>
</tbody>
</table>

Table 1-3. Range of bending (radius measured) deformation for peripheral arteries.
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<th></th>
<th>Unstented Arteries</th>
<th>Stented Arteries</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Distal SFA Proximal Popliteal Artery</td>
<td>Distal SFA Proximal Popliteal Artery</td>
</tr>
<tr>
<td></td>
<td>Middle SFA</td>
<td>Middle SFA</td>
</tr>
<tr>
<td></td>
<td>Popliteal Artery (90/90 Knee/ Hip Flexion)</td>
<td>Popliteal Artery (90/90 Knee/ Hip Flexion)</td>
</tr>
<tr>
<td>Torsion (deg/cm)</td>
<td>1.3 ± 0.8%</td>
<td>1.8 ± 1.1%</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2.1 ± 1.3%</td>
</tr>
<tr>
<td>Torsion (length unknown)</td>
<td>60 ± 34 deg</td>
<td>None Detected</td>
</tr>
</tbody>
</table>

Table 1-4. Range of torsional (angular and percent) deformation for popliteal arteries.

The figures above illustrate studies which investigated the associated metrics. It also suggests that there may be more data needed for how stented arteries react in torsion as this information is absent in the chart.

The carotid artery was described by two regions, the internal carotid artery (ICA) and the common carotid artery (CCA). The primary modes of strain were bending and torsion. These values differed greatly between the two regions and because of the large values, angulation was noted as a concern with stented patients. There is the potential that the radial contraction of the carotid artery due to swallowing, though less severe, occurs more frequently and thus could potentially result in fatigue failure.
<table>
<thead>
<tr>
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<th>Unstented Arteries</th>
<th>Stented Arteries</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Internal Carotid</td>
<td>Common Carotid</td>
</tr>
<tr>
<td>Torsion Left</td>
<td>15.48 (13.68–53.78)</td>
<td>34.38 (20.48–82.58)</td>
</tr>
<tr>
<td>Torsion Right</td>
<td>28.68 (5.28–65.48)</td>
<td>26.98 (14.38–38.18)</td>
</tr>
<tr>
<td>Artery Max</td>
<td>65.48 deg</td>
<td>82.58 deg</td>
</tr>
<tr>
<td>Mode Max</td>
<td>82.58 deg</td>
<td>69.48 deg</td>
</tr>
<tr>
<td>Overall Max</td>
<td></td>
<td>82.58 deg</td>
</tr>
</tbody>
</table>

Table 1-5. Range of torsional deformation for carotid arteries.

In general, the magnitude and mode of artery deformation between test subjects was highly variable. However, within the subject, the range of motion and deformation of arteries was largely symmetric. Unstented arteries were often compared with stented arteries, which verified the assertion that the stents are less yielding than the artery alone. Suggested fatigue tests were run at 7-10 Hz for 1 million or more cycles. However, a combined modes of strain test was not found amongst the data. For the fatigue tests, stents were suspended in a flexible silicon tubing which was found to most accurately emulate the properties of an artery. No tests had reported using pulsatile fluid flow in their experiment, yet it was noted as a potential further step.

1.2.2 State of the Art in Stent Fatigue Testing

Though there are multiple modeling approaches, there exist multiple analysis methods as well, serving separate purposes. A primary concern with stents is the fatigue failure which can produce a fatal outcome. Subsequently, substantial research efforts have been made to address the reliability concerns of stents in endurance cycles. As found in Pelton et al. 2003, a diamond shaped Nitinol sample was modeled and subjected
to 10 million cycles in simulation. Using a non-linear fatigue model for Nitinol, the authors were able to validate the fatigue life predicted by the FEM approach [19]. Due to the dynamic nature of the vascular environment, it is extremely difficult to quantize a life cycle let alone create prototypes, fixtures and run tests. Typically these analyses are performed as a precursor to, or in conjunction with physical testing methods.

Lowe notes that the increasing rate of in-stent restenosis occurrences can be approached from a standpoint of treatment or prevention. Current methods of treatment such as Percutaneous Transluminal Coronary Angioplasty, Directional Coronary Atherectomy and High Speed Rotational Atherectomy, have shown improvement but lead to further reoccurrences. Another approach involving radiation treatment is delicate and not well understood. Prevention techniques using polymer coatings have thus far yielded the greatest efficacy, however questions remain regarding the ideal chemicals and acceptable toxicity. Conventional drug delivery systems are also under investigation, with a gene transfer treatment being the most successful, though in very early stages. Localized delivery of any treatment or preventive mode is preferable for optimal results [20].

In their research, Auricchio points to the importance of understanding shape memory alloys due to their increasing presence in medical devices, specifically stents. A parametric formula called the Free Energy Density formula, $\Psi(\Theta,e,T,\text{etr})$ (pg.227), is presented, which aims to improve FEA analysis of stents during deployment. The author also illustrates the critical characteristics of an accurate simulation such as: accurate modeling of peripheral artery and stent design; simulation of stent manufacturing process; accurate material properties (stent and arterial vessel); accurate model of stent
deployment; and accurate loading simulations. In the analysis, some of these models are simplified by assumptions (p.234). The author claims to experimentally calibrate SMA models based upon strain-temperature curves at varied constant stresses. Additionally, it is stated that the deployment simulation in Abaqus is successful. Upon inspection, it appears plausible, however an experimental comparison is not provided for verification. Further research into this area is encouraged [21].

This case study details a medical approach to coronary restenosis due to fracture. “The fracture was confirmed with both conventional coronary angiography and MSCT which had higher sensitivity and specificity for detection of stent fractures in previous studies [17]. In our case MSCT did not show coronary aneurysm which was clearly verified with coronary angiography.” The results support the cited research, stating MSCT is the preferred imaging technique to identify stent fractures post implementation [22].

The case study outlines the chain of events, beginning with PTCA installed drug eluting stent. Within hours, left main coronary thrombosis was detected as well as myocardial necrosis. The authors note that the imaging mode utilized was transesophageal echocardiography which is standard for “visualizing coronary arteries” however, “is the first case of a left main stent thrombosis [being] identified” using this imaging technique. An established method for identifying stent fractures in vivo is intravascular ultrasound (IVUS) [23].

Canan and Lee performed an extensive review in order to determine the frequency of fracture in drug eluting stents corresponding to coronary arteries. A retrospective investigation was performed to identify pertinent research and merge the results. The
research found was from a single institution with a defined method thus creating a heterogeneous perspective. The summation of their findings correlated increased stent angulation, bending and length to an increased fracture rate, particularly when said angle is greater than 45 degrees. Overlapping stents also posed very high risk for stent fracture. The investigation also noted that stents with a larger diameter tended to be less susceptible to fracture. Additionally, Cypher (closed cell) stents are assumed to have higher incidence of fracture than Taxus (open cell). However due to imaging difficulties, the observability of fractures is lower in the latter case so a definitive conclusion cannot be made without a head-to-head comparison [24].

A baseline comparison study was performed to evaluate the relative effectiveness of intravascular ultrasound (IVUS) and serial angiography imaging modalities in identifying stent failure in the coronary artery. One hundred and two sirolimus-eluting stents (SES) were placed into 56 patients. Baseline images were taken using both techniques as well as during subsequent follow-ups at six and twelve months. Stent failure was categorized among three types, including; stent fracture, hinge movement and disassociation. For all three failure modes, IVUS provided earlier and more complete recognition when compared with serial angiography. However, further research is needed to increase the sample size. Additionally, the study assumes IVUS identified 100% of fractures which is not likely. Because IVUS is not a standard procedure for imaging coronary stents, SES failure and degradation is likely under-diagnosed [25].

Stents are subjected to a series of tests in their design phase, often incorporating cyclic loading to simulate fatigue stresses. Upon completion of a predetermined cycle count, the stent is examined for fractures. Several methods are used for In Situ fracture
detection, including visual inspection. This involves placing a test specimen under a microscope and visually identify any fractures. This is a very tedious and time consuming approach. Additionally, it is difficult if not impossible to determine precisely when the fracture occurred. The assessment becomes qualitative, allowing only for a 'pass' or 'fail.' This approach is also known as "test-to-success" and has been criticized for the lack of information and insight which accompanies it.

![Figure 1-2. Fracture detection by visual inspection using a digital microscope.](image)

Fracture detection by visual inspection using a digital microscope is a delicate and time consuming procedure. It can only be performed once the test has been completed. Some methods allow for the test specimens to remain in testing position allowing for intermittent evaluations. Yet the time resolution is not adequate for generating a reliable FEA correlation. Additionally, the evaluation techniques do not provide the resolution or sensitivity necessary to make accurate judgments about fractures.
Table 1-6. Evaluation of In-situ methods. Highlighted methods were merged for use in fracture detection.

Another method of fracture detection is described in A. Mehta, et al. where a technique called X-ray microdiffraction allows researchers to view the material at the grain structure level as well as observe the specific phase state, including austenitic and martensitic phases [26]. This method is useful for observing phase change in fatigue and failure, however it's range of view is very limited. For this reason, it is less useful as a dynamic observation tool where the fracture location is not already known. The primary application of this technology thus far has only been to study fracture propagation in bare metal at a known location and not to specifically identify a fracture within an entire stent.
A method to determine the specific failure time and location in a stent during a fatigue testing cycle would yield improved understanding of stent weaknesses. Additionally, the results can be correlated to finite element analyses and eventually performed as simulations, avoiding expensive and time consuming physical testing. This development could also offer a wide range of benefits including; better iterative stent design; stent customization; and stress monitoring during deployment. Each of these objectives contribute to the global goal of improving patient care.
1.2.3 Review of Stent Designs

The term "stent" is not exclusive to circulatory system applications as it can be used in the respiratory system (i.e. esophageal stent), excretory systems (i.e. uretal stent) among others. However, for the purpose of this discussion, stents will reference vascular applications.

Stents have a variety of different configurations that affect everything from deployment procedures to overall efficacy. These parameters include material type, construction geometry and surface treatment and/or coating. The differentiating features of these modes as well as their general impact on success will be briefly discussed.

Stent designs have incorporated a variety of materials including metals, alloys, and various polymers. Appropriate material selection for a stent is a critical factor for a several of the following reasons. A standard deployment techniques involves fluoroscopy which necessitates that the stent have sufficient radiopacity. This characteristic is different among materials. Additionally, biocompatibility is very important not only for the health and longevity of the device but for physiological reaction to foreign objects. A third metric is the flexibility of the material. Though coronary stents require less flexibility on account of their limited deformation, peripheral arteries must endure high deformations as illustrated in tables 1-1 through 1-5.

Metals: Initial stent development was based around stainless steel due to its biocompatibility, however radiopacity nor flexibility were ideal. Later, gold stents were developed because of their well known visibility in the fluoroscopic spectrum. Additionally, they exhibited similar inert qualities as stainless steel, and drastically improved flexibility. However, gold proved excessively expensive which eventually led
to hybrid designs such as Medtronic's Bestent, a stainless steel mesh with radiopaque gold markers located at the distal ends of the device [27]. A less frequently used metal, Tantalum provides ideal radiopacity and excellent ductility. Though it exhibits high corrosion resistance, it does not make a very good biocompatible material [28]. Examples of its implementation include Wiktor Stent by Medtronic and the Tantalum Cordis Stent by Cordis Corporation [29].

Advanced Alloys: Though not the only advanced alloy used in stent design, Nickel-Titanium, more commonly known as "Nitinol" or "NiTi" is the most prevalent in recent years. It is not as visible during fluoroscopy, but does have the advantage of being a shape memory alloy (SMA). These unique metals present unusual, but highly advantageous flexibility characteristics. Additionally, it has the ability to return to its original structural composition if plastically deformed by heating it above its phase transformation temperature. Composed of 54.5-57.0% Nickel with the remaining balance being Titanium [30], Nitinol has been widely used in the biomedical industry because of its superb biocompatible qualities, especially when proper surface treatment is employed [31, 32].

Polymers: Similar to Nitinol, polymer stents do not inherently exhibit strong radiopacity. However, with many of the polymeric based materials, fluorescent additives can be used to supplement these deficiencies. Where supplements cannot be used, often metallic based markers can be placed at the distal ends of the stent. There is a significant number of polymer based approaches, including silicone; polyethylene; polyurethane to name a few. Though some can be characterized as biocompatible, others can cause a layer of protein and biofilm to aggregate locally and present potential restenosis risks. A
newer form of polymeric stents involves biodegradable and/or bioabsorbable materials. Though structural integrity is less of concern due to the intended degradation/absorption, it is still a matter of investigation and optimization as dislocated segments can be hazardous to vascular health. In contrast, polymer based stents have a strong ability to localize delivery of drugs by incorporating or being composed of medication.

Construction geometry is also vital design consideration for stents. As illustrated previously, vascular tissue endures extreme deformations. Though selecting a flexible material aids in the reduction of stress accumulation, geometry must also supplement this objective. Stent geometry is partially driven by the selected manufacturing approach, several of which include: laser cut tube; welded wire; knitted wire; braided wire; wire coil; and flat wire coil [33]. Certain designs tolerate deformation more than others, however this is a significant focus of current research in the vascular stenting. Because fatigue stress accumulation is poorly understood in these complex structures, much investigation remains. Additionally, the contribution of this research aims to contribute to this body of knowledge.
As shown by Stoeckel et al (2003), the top image displays a severely corroded Nitinol explant after five months of exposure, a result of poor surface finish. The explant below it was electropolished and appears in excellent condition after 12 months. Though structural integrity is a concern with stents, the immediate and short term efficacy is also of critical importance. The stent must not only allow for the expansion of the vascular tissue, but also ensure that it does not provide a foundation for new plaque deposits, or reactionary growth. This has led to the development of drug eluting stents (DES) which are coated with special medication. The medication slowly releases chemicals which prevent plaque accumulation and also encourages adoption by the surrounding tissue. Surface finish is also a substantial factor in the longevity of most metallic and alloyed stents. Poor surface finish results in unacceptable porosity and excessive corrosion exposure [33].
Chapter 2: Stent Failure Modes

Because stents are essentially a flexible scaffolding, they must endure similar deformations which are encountered by their encompassing vasculature. Artery dynamics can be very extreme as illustrated by Smouse et al. (2005) [6]. These extreme deformations can be most noticeably observed in the stented femoropopliteal artery (FPA) Segment under 90/90 knee/hip flexion which endures up to 69° bending, compression strain of 25%. Additionally, the proximal popliteal artery (PPA) experiences up to a maximum of 29° bending and 9% of compression strain, while internal carotid artery (ICA) experiences up to 69.48° of torsion and 52.2° of bending [8].

With the surging popularity of stenting, stent design and development have correspondingly become more complex and rigorous. Though due to the wide range of physiological conditions, no singular design has yet to adequately satisfy a general patient population. Moreover, even optimized designs using shape memory alloys (SMA's) are susceptible to high failure rates on account of the extreme loading conditions as well as hyper-deployment.

2.1 Generalized Failure Mechanisms

Failure modes generally involve similar mechanisms. Over the past decade, several researchers have established a classification system for stent fracture. The common approach defines any strand in the stent as a "strut." There is generally no differentiation between location, length width or any other analogous factors. As well, the categories make some mention of single and multiple fractures, though further detail can be included. Other modes that are not always accounted for involve the buckling and transection phenomenon.
Nair and Quadros (2011) aggregated current classification schemes for stent fractures [34]. As is illustrated, there has yet to be a defining standard. For the purpose of this research, the classification system presented by Allie et al (2004) is most convenient [35, 36]. A single strut fracture (Type I) typically accelerates local fracturing due to the imbalance of residual stresses. Even if a single fracture does not generate successive fractures, it can pierce tissue and plaque or serve as collection point for particulates. These create a substantial risk factor for restenosis and require follow-up treatment if discovered.
A Type II fracture occurs when there are two or more fractures. If they are not in adjacent sites, neither is considered a propagation of the other. However, if they are adjacent to one another it may indicate that similar stresses contributed to each occurrence. A Type III fracture is a very serious situation. Successive fractures have separated the stent into two or more pieces. In this stage, the pieces of the stent have had no relative movement to the vascular tissue. In Type IV fracture, movement would have occurred and could prove lethal if not promptly addressed. Allie et al (2004) did not classify a Type V fracture as is given in Nair and Quadros.

![Figure 2-2. In Vivo angiographic images of the various types of stent fracture.](image)

The frequency, location and severity of these fracture types is dependent on a variety of conditions including deployment procedure, patient activity, stent dimensions and mechanical properties of the patient vasculature. Because of this, there is a substantial demand for understanding the influence of parameters on the failure mechanisms of stents.

2.2 In-situ Experimentally Observed Stent Failures

Because fractures modes are highly dependent upon the structure, design and materials in the stent, it is important to thoroughly understand the specific failure modes of a particular stent. For the associated research, a Protégé Everflex stent was selected. The stent geometry incorporates a linear pattern of rings with a saw-tooth profile. Each successive ring is connected to the previous via four joints which will be termed "connectors" in subsequent discussion. The connector junction is located at the convergence point of four struts, two leading in from the proximal ring, and two leading out from the distal ring. This is with the exception that the first ring and last ring have half connectors at their start and end points, respectively. The individual rings are composed of 32 nearly linear segments which will be referred to as "struts." The wishbone shape seen below is termed a "v-strut" and is technically two struts. Though each ring has the same geometry as the one before and after it, the connector locations propagate helically through the stent. Illustrations of the aforementioned features can be observed below.
Figure 2-3. High resolution images used to reverse engineer and create a computer model from an existing device.

Figure 2-4. The steps involved in creating a solid model for numerical simulation are shown.

As can be seen in the previous figure, strut features are not all uniform. Though the struts themselves are generally consistent in shape, the geometrical features are not the same. This is particularly important with regard to the proposed fracture detection
method. When a strut fractures, it is characterized as the failure of a feature. For the purpose of this research, identifiable fractures are complete fractures.

A single strut fracture will either occur along the "long-strut" feature or the "short-strut" feature on the Protégé Everflex. A long-strut fracture is generally rare because stress concentrations typically do not accumulate quickly in this region (Avdeev et al.) [38]. This is likely the result of having a longer segment capable of distributing strain over a greater length. The short-strut feature is less immune to fracture by the same reasoning.

Figure 2-5. The affects of fatigue testing on a stent.

The stent on left is illustrates fracture as a result of fatigue cycling. The stent on the right illustrates the propagation of the initial fracture from the left, demonstrating the pathogenic process of single strut fractures. During the procedure, fractures can accumulate. Though this is known to occur In Vivo, there is value in performing analyses immediately following the initial fracture.
2.3 Non-Linearities in Material Modeling of Shape Memory Alloys

As a result of the increasing popularity of shape memory alloys in medical devices, simulating it's performance characteristics has become ever more necessary. Though SMA's such as Nitinol don't adhere to the traditional material modeling scheme where the stress-strain curve is largely linear. This poses a tremendous challenge when performing numerical analysis. A unique mathematical model must be introduced to account for additional parameters that wouldn't typically change the performance of the material as drastically.

\[
(\sigma - \sigma_0) = C \cdot (\varepsilon - \varepsilon_0) + \Theta(T - T_0) + \Omega(\xi - \xi_0)
\]  

(2-1)

The formula above illustrates the input variables required for accurately determining stress states in shape memory alloys. The SMA material model, where \( \sigma \) is the Piola-Kirchhoff stress tensor. \( C \) is the stiffness tensor. \( \varepsilon \) is the Green-St. Venant strain tensor. \( \Theta \) is the thermoelastic tensor, \( T \) is the temperature. \( \Omega \) is the metallurgical transformation tensor. \( \xi \) is the martensite volume fraction, and variables with the subscript "o" refer to the initial conditions where the material properties are constant [38].

2.4 Photogrammetry and Fatigue Testing

Previous research has used photogrammetry to closely emulate stent testing conditions in a simulation environment. Shams et al. (2011) has shown that this approach can accurately identify stress concentrations in the stent and by dynamically adjusting boundary conditions [39]. If the temporal and physical boundary conditions can be
accurately defined for simulation, based on experimental results, a thorough
understanding of fatigue and stress accumulation in stents can be achieved.

Figure 2-6. Illustration of reduced order model and associated von mises stress distribution results, given experimental boundary conditions.

The experimental instrumentation known as the Peripheral Artery Stent Testing Apparatus (PASTA) was developed to cyclically apply three forms of strain on the subject stent, including bending, tension/compression and torsion. Each of these is individually controlled and capable of frequencies as high as 3Hz. Each deformation is intended to closely emulate the maximum respective deformation from the aforementioned aggregation of clinical studies.
The Peripheral Artery Stent Testing Apparatus (PASTA) is capable of inducing three forms of biomechanically accurate deformation and uses a CCD camera to observe the deformation response in the stent. As a the Protégé Everflex Nitinol stent undergoes a deformation the photogrammetric process (illustrated below) establishes the dynamic boundary conditions for the ANSYS simulation. Extracted boundary conditions from PASTA are subsequently input to a reduced order model of the stent for analysis. According to Avdeev and Shams (2010), the reduced order model expedites computation while maintaining quality of results [38]. Such an approach provides computational efficiency, a significant factor in fatigue analysis.
Figure 2-8. Stent images through the course of the photogrammetric process

The photogrammetric process and associated analysis is one tool which aids in the investigation stent fractures. However, the results are limited by the available information. Providing greater detail on fractures in the temporal domain, as well automating the location detection has the potential to scale and expedite stent analysis, thereby improving design and efficacy.
CHAPTER 3: FRACTURE DETECTION IN BARE METAL STENTS

3.1 Theoretical Approach

3.1.1 Lumped Resistance Model

The stent is constructed from many interconnected struts in a specific pattern. The stent can be modeled as a network of lumped resistive elements. Each lumped element represents either a strut or a connector with its own resistance value. Any fracture of a strut or a connector affects the network resistance. Any deviation from the reference resistance value (unfractured stent) can be measured using a low resistance ohm-meter.

![Image of network resistance scheme](image)

Figure 3-1. Representation of the network resistance scheme of the Protégé Everflex stent.

\[
R_{\text{Stent}} = \left[ \frac{R_{\text{SS}} \cdot R_{\text{LS}}}{R_{\text{SS}} \cdot n_{\text{LS}} + R_{\text{LS}} \cdot n_{\text{SS}}} \right] \cdot N_{\text{Rings}} \tag{3-1}
\]

The theoretical model above gives the effective resistance of the subject stent where: \( R_{\text{Stent}} \) is resistance of an unfractured stent (ohms); \( R_{SS} \) is resistance of a short strut...
(ohms); $R_{LS}$ is resistance of a long strut (ohms); $n_{SS}$ is the number of short struts in a ring; $n_{LS}$ is the number of long struts in a ring; and $N_{Rings}$ is the number of rings in the stent itself.

Topological network changes caused by one or several fractures will be indicative of the type of fracture which has occurred. Using the grading scale from Nakazawa, et al. 2009, fractures modes are categorized. Current theoretical models only account for Grade I and one form of Grade II (double connector fracture), however additional models are currently in progress. Depending on the fracture location within in a stent ring, a different affect on the resistance can be had. The theoretical model does account for this.

Theoretical modeling attempts to predict the relative difference in resistivity that would occur given a particular fractures type. Considering the Everflex Protégé Nitinol stent, there are four specific features that can be fractured, including the short strut, the long strut, connector and the two connector fractures (Grade II).

Initial theoretical results were left in a unitless form, allowing for the percent difference in resistance to be calculated rather than an explicit form of resistance. Though the stent is a pattern of struts, the strut itself is not uniform in contour, making it difficult to calculate the resistance during an initial investigation. However, by understanding that the strut has some resistance value, "R", a relative difference in resistance can be determined.

3.1.2 Joule Heat Model

Resistive elements have a thermal response due to current flow, known as the Joule Heat Effect. If current is applied to a fractured stent, the fracture location can be
identified by using a high resolution thermal imaging device and tracing the thermal gradient to the cold zone.

In order to establish a location of a fracture, the Joule effect is utilized, though must be coupled with the free convection model to achieve a steady state result. The theoretical Joule Heat model coupled with free convection. Following that is the coupled model in simplified format.

\[ Q = \left[ \sum_{i=1}^{n_{SS}} (I_{SS} \times R_{SS}) + \sum_{i=1}^{n_{LS}} (I_{LS} \times R_{LS}) \right]^2 - h \times A(T - T_\infty) \quad (3-2) \]

\[ Q = I^2R - h \times A(\Delta T) \quad (3-3) \]

In a complete stent, there should be current flow of varying degrees (based on resistance) through all struts. However, it can be observed that a completely fractured strut will have no current flow, thus having no heat generation. What's more, two stents given the same current load, the fractured stent will experience localized regions of greater current density as it compensates for a missing strut. These regions would be expected to experience higher heat generation than they would in an unfractured stent.

### 3.2 Numerical Approach

Numerical models are particularly useful in non-uniform multiphysics problems or where complicated geometry must be analyzed. Stents serve as a prime example for both of these cases. Moreover, a numerical model is an important step for validating theoretical models and calibrating experimental boundary conditions. In this particular case, it also provides additional insight for the current flow path and voltage potential
regions. These qualities would be difficult or impossible to visualize with any other method.

3.2.1 Discrete Model

A discrete parametric model of the stent was generated in ANSYS Multiphysics environment. This model was modified from a mixed model (Solid186, Beam188 & Solsh190) to a 3-D 20-Node Coupled-Field Solid (Solid226). The element has twenty nodes with up to five degrees of freedom per node. Structural capabilities are elastic only and include large deflection and stress stiffening. Thermoelectric capabilities include Seebeck, Peltier, and Thomson effects, as well as Joule heating.

Figure 3-2. An illustration of Solid226 element type. This particular element allows for thermal-electric analysis to be conducted.
3.2.2 Setup and Assumptions

The approach for theoretical analysis of stent resistance can vary based upon the stent design and material. The method developed in this research is specific to the Protégé Everflex model however can be modified for other stents. Several critical assumptions are made, including:

1) steady state analysis
2) external temperature, $T_\infty = 295$°K (RT)
3) all struts have equivalent resistance
4) electrical resistance is constant (small $\Delta T$)
5) no forced convection
6) heat conduction propagates within stent
7) no radiation heat loss
8) no phase transformation
Once the dimensional model is created in ANSYS, material properties and parameters must be provided. The specification for electrical resistivity of Nitinol (SE 508 wire) were extracted from the manufacturers documentation. This value being 8.2mΩ/mm at room temperature. Additionally, the thermal conductivity was value was obtained from the same literature, 8.6 W/m°K.

Free convection must be accounted for in the numerical analysis or the simulation will not converge. This is because without a balancing term, the Joule Heat model will effectively heat the stent to infinity. In the numerical model, a free convection condition was applied to all surface nodes using the free convection coefficient value range determined by Incropera and DeWitt (1984), $9 - 25 \times 10^{-6}$ (J/mm²s°K) [40]. Using the highest value of this range, $25 \times 10^{-6}$ (J/mm²s°K), will account for the most severe cooling effect, Bhattacharyya et al. (1995) [41]. This would effectively tighten the temperature gradient in the results and show the minimum sensitivity to fracture location, or least detectable change scenario. The temperature associated with this value is, $T_\infty = 295°K$ (RT).

Figure 3-4. A computer model of the Nitinol stent is created for numerical analysis.
Though the manufacturing process of the Protégé Everflex is conducted by laser cutting a Nitinol tube with a specific pattern, the numerical model is generated via patterned struts which create a ring. The rings are subsequently patterned in the Z-direction to complete the stent.

![Figure 3-5](image1.png)

Figure 3-5. A ring of a Protégé Everflex stent is modeled in ANSYS and composed of patterned struts. Rings are connected in series to create a complete stent.

![Figure 3-6](image2.png)

Figure 3-6. A complete stent is composed of a pattern of rings which are generated by a pattern of struts.

### 3.3 Experimental Approach

#### 3.3.1 Sample Selection

An unfractured Protégé Everflex stent with a 10mm diameter is used for the experimental portion of the study. Due to numerical computation expenses, an analysis was chosen to be performed on 15 ringed models. Hence the experimental procedure
dictates that electrical loading must be applied across only a section of stent 15 rings long.

### 3.3.2 Resistance Measurement

Two stainless steel probes were designed such that contact is made radially about the stent at a specific distance along the length of the stent. This is to ensure consistent electrical contact is maintained throughout the testing process. The probes are located at the distal and proximal ends of the stent, precisely 15 rings apart. The stent is in a free state with no mode of deformation applied to it. This setup is considered a 'lumped resistive unit' (LRU). The proximal and distal ends being the positive (+) and negative (−) terminals respectively.

![Figure 3-7. Electrical loading of 15 rings in a Protégé Everflex Nitinol stent.](image)

Resistance measurements are collected using DC regulated power supply [BK Precision 1665]. A specific current (100 mA) is applied and the voltage is recorded. Using a form of Ohm's Law, the effective resistance can be determined. Below, Ohms law, used to calculate total resistance across 15 rings of a Nitinol stent.

\[
\sum R = \frac{V(x) - V(0)}{\sum I}
\]  

(4)
The LRU resistance is initially measured without the stent as a calibration check. Each stent is placed in the LRU fixture prior to its respective fracture occurrence. The resistance is measured and logged.

### 3.3.3 Thermal Measurement

After each round of resistance measurements a thermal image must be obtained of the stent. Ambient air temperature will be 22 ± 1°C. An enclosure provides a closed environment for the stent during heating, so as to ensure no moving air can affect the convection process. Additionally, the enclosure will serve to mitigate any changing external radiating bodies that could interfere with data collection. A constant current load of 100 mA is applied to the fixture. A thermal imager mounted on a track which can be repositioned precisely will have the center of the stent as the center point of focus. The thermal imager monitors the temperature profile of the stent until it exits the transient stage and becomes steady state. An image is subsequently captured. Two more thermal images are taken which will be merged later to reduce any random effects.
Figure 3-8. Experimental setup for thermal and electrical evaluation.

The setup above incorporates many features which must remain consistent through the course of experiments. The chamber's walls though transparent in the visual spectrum, baffle external thermal bodies such as humans, lights, computers and monitors among many others. Additionally, it impedes airflow creating an insulated environment, allowing air temperature to be accurately monitored and regulated.
Chapter 4: Fracture Detection and Location Results

4.1 Fracture Detection by Resistance

Identifying a stent fracture by detecting a change in resistance has the potential to provide a non-invasive method for determining lifespan of stents in cyclic loading conditions. Various approaches were taken to evaluate the feasibility of this approach. Initially the theoretical modeling was conducted, followed by numerical simulation and finally concluded with an experiment. The following illustrates the results achieved and a thorough analysis of the data.

4.1.1 Theoretical Model

As described previously, a quantitative resistance model is not currently feasible. However, by assuming the stent to be a series of patterned individual struts with equivalent resistance, a unitless model can be developed to indicate sensitivity to fractures. Equation 3-1 from 3.1.1 was programmed in Matlab to provide a baseline number for an unfractured stent containing 15 rings. A subsequent implementation of that equation was performed with a short strut, long strut or connector(s) removed.
Figure 4-1. Stent features and corresponding fracture locations.
Figure 4-2. Relative change in resistance as a result of fracture (theoretical model).

A model was developed for the Protégé Everflex geometrical pattern which approximates the relative increase in resistance given several types of fracture scenarios. The features which would provide the greatest path for resistance in the device inversely have the least affect on overall resistance. However this model does not account for any non-linearities in resistance which could be induced by the joule heating. To investigate this phenomenon, a numerical analysis should be performed.

4.1.2 Numerical Simulation

An idealized numerical model was constructed using a reverse engineering techniques. The geometry was patterned in ANSYS using solid elements with thermal electric elements (SOLID 226). Material properties were obtained for RT conditions (22°C). Thermal conductivity within the stent is accounted for and free convection is assumed using a free convection coefficient for wire, determined by Incropera and DeWitt (1984) [40].
Resistance is indirectly calculated based on the effective voltage potential between the proximal and distal ends of the stents. As previously described, this voltage value is divided by 100 mA (0.1 A) to determine the true resistance. Fractured resistive values can be compared to the unfractured resistance in order to provide a comparison metric for the theoretical model.

![Figure 4-3](image1.png)  
**Figure 4-3.** Voltage potential within an unfractured stent from end to end. The maximum value is used for determining effective resistance.

![Figure 4-4](image2.png)  
**Figure 4-4.** Voltage potential within a fractured stent (double connector) from end to end. Note that the result provides little visible indication that a fracture has occurred.
The examples above demonstrate the visual output generated by the simulation. The voltage potential values are extracted and used to calculate the resistance. A mesh sensitivity analysis was performed to optimize accuracy and computational expense. Further refinement by a factor of two resulted in ~ 1/1000\textsuperscript{th} % difference with roughly a tenfold increase in computation time.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Voltage</th>
<th>Resistance (Ohms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>No Fracture</td>
<td>1.08</td>
<td>10.78</td>
</tr>
<tr>
<td>Long Strut</td>
<td>1.09</td>
<td>10.9</td>
</tr>
<tr>
<td>Short Strut</td>
<td>1.11</td>
<td>11.05</td>
</tr>
<tr>
<td>Connector</td>
<td>1.14</td>
<td>11.39</td>
</tr>
<tr>
<td>Double Connector</td>
<td>1.22</td>
<td>12.24</td>
</tr>
</tbody>
</table>

Table 4-1. Lumped resistance as solved by numerical simulation.

Figure 4-5. The relative increase in resistance for the associated fracture types (numerical simulation).

The numerical simulation also follows the expected trend for fracture types. The concept is further supported, however the experiment must be conducted for any
successful validation. Moreover, the results will be necessary to understand any correlation between real world and simulation environments. As discussed previously, there is substantial design value in identifying such relationships.

4.1.3 Experimental Results

The experimental setup for resistive testing was combined with the thermal testing. Fifteen rings of the Nitinol stent (Protégé Everflex) were used to bridge two aluminum cylinders. The effective resistance of the testing unit was evaluated at 1 ohm. This was correspondingly subtracted from the results, 0.1 volt per 0.1 amp, so as to provide true resistance.

Figure 4-6. The stent is suspended over a gap to match the numerical simulation boundary conditions as closely as possible.

To evaluate the linearity of resistance in each scenario, amperage was increased by increments of 100mA and voltage was recorded. To account for any hysteresis effects, the values were recorded as amperage descended from the maximum value (>0.7 amps) and averaged. Values were often identical and never significantly different. This is a positive indication that the experiment was stable within its own setup.
### Electrical Resistance of Stent Based on Experimental Results

*(100mA Applied Current)*

<table>
<thead>
<tr>
<th>Condition</th>
<th>Voltage</th>
<th>Resistance (Ohms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>No Fracture</td>
<td>0.11 - 0.20</td>
<td>1.1 - 2.0</td>
</tr>
<tr>
<td>Long Strut</td>
<td>0.16</td>
<td>1.6</td>
</tr>
<tr>
<td>Short Strut</td>
<td>0.32</td>
<td>3.2</td>
</tr>
<tr>
<td>Connector</td>
<td>0.50</td>
<td>5.0</td>
</tr>
<tr>
<td>Double Connector</td>
<td>0.36</td>
<td>3.6</td>
</tr>
</tbody>
</table>

Table 4-2. Lumped resistance as determined by experimental procedure.

### Increase in Resistance by Fracture Type

Experimental Results

- Long Strut: 45.45%
- Short Strut: 146.15%
- Connector: 150.00%
- Double Connector: 80.00%

Figure 4-7. Relative increase in resistance from non-fracture to fractured state as determined by experimental procedure.
Figure 4-8. Long strut fracture vs. original non-fractured state (experimental data).

Figure 4-9. Short strut fracture vs. original non-fractured state (experimental data).
Figure 4-10. Connector strut fracture vs. original non-fractured state (experimental data).

Figure 4-11. Double connector strut fracture vs. original non-fractured state (experimental data).
4.1.4 Discussion

Upon evaluating the resistive performance of the stents against the theoretical model and numerical simulations, a significant gaps exists. The theory and simulation do exhibit the predicted tendency to increase in resistance as the severity of the fracture type was increased. There was roughly a factor of two difference between "change in resistance" values, could be explained by an oversimplification of the strut modeling and corresponding features. Additionally, the theoretical model assumes linear progression of current in a 2-D circuit, whereas the simulation contains helical propagation in 3-D space. Lastly, the simulation accounted for the joule heating effect. The resistive properties are subject to change as function of several parameters driven by temperature.

The experiment exhibited a significantly higher sensitivity to fracture than was anticipated. Though a positive indication for fracture detection, the inconsistent response is difficult to understand. Novák et al (2007) cite the lack of understanding in electric resistivity during thermomechanical testing as a primary reason why Nitinol and other SMA's are not yet suitable for detection devices [42]. This is because Nitinol can exhibit non-linear resistive response during phase transformation, and even develop a negative slope (higher temperature reduces resistance) during the R-phase where crystalline structures become better aligned. This could explain the unusual drop-off in the "double connector fracture" experiment at 0.5 Amps.

An additional explanation for the wide discrepancies in resistance could be that the reverse-engineered model did not exactly replicate the physical design. Because of the complicated structure, smaller portions were patterned to generate a representation of
the stent. Inconsistencies in any manufacturing process, so certainly the model was not exact replica, only as close as could be approximated.

4.2 Detecting Fracture Location by Infrared Imaging

Detecting fracture occurrence is the first step to improving understanding of fatigue life of stents. However, identifying the location and type in a noninvasive and seamless manner would provide substantial information on initial modes while maintaining the ability to observe how failure propagates. Thermal imaging an electrically heated stent has the potential to serve each of those functions.

4.2.1 Numerical Simulation

Numerical simulations were performed using developed user routines written in ANSYS APDL command language. Boundary- and initial conditions were similar to the experimental environment. The coupled-field finite element model consisted of 692,220 nodes. The material properties used for simulations are summarized in Table.

<table>
<thead>
<tr>
<th>Properties</th>
<th>Values</th>
<th>Units</th>
<th>Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thermal Conductivity</td>
<td>8.60E-03</td>
<td>W/m*C</td>
<td>Nitinol Corporation (NDC)</td>
</tr>
<tr>
<td>Electrical Resistivity</td>
<td>8.20E-03</td>
<td>mΩ*mm</td>
<td>Nitinol Corporation (NDC)</td>
</tr>
</tbody>
</table>

Table 4-3. Material properties specific to Nitinol used in the numerical simulation.
The figure above illustrates the divergence of temperature extremes. This would indicate that it may be possible to identify a fracture by observing a change in the local minimum and maximum temperature values. However, the primary objective is to locate the region of the fracture, not simply maximum or minimum of temperature field. By evaluating this chart, it could be observed that the standard deviation of temperatures is likely to increase with fracture severity. Future work may consider this an important metric. Results of temperature mapping can be observed in the following figure.
Figure 4-13. The resultant current density map illustrates the current propagation (100 mA applied current). The correlation between current density and Joule Heating can be observed.
The unfractured stent exhibits a fairly uniform distribution of current, but it does helically propagate through the stent. This can be explained by the non-symmetric properties about the connector junction. One side of the junction has three struts between it and the next junction, whereas the other side has 5 struts between. All struts assumed equal, the additive property of series resistance indicates the three strut section will have lowered resistance. Therefore, current flow will be biased towards the path of least resistance and be greater in the three strut segment. Following the Joule Heat law, this will generate greater heat flux. This is easily observed in the figure, and provides visual indication as to the location of the fracture. Testing is essential to validate or refute the simulation.

4.2.2 Experimental Results

The experimental setup consists of a carefully placed thermal imaging unit. The stent itself is in a bracket which is suspended in air so as to reduce any heat accumulation. This also allows for free convection, an environmental condition given in the numerical simulation. The bracket uses a corner locating feature so that the stent can be moved (so as to manually induce a fracture), yet return the stent to the original position. This is critical as the image processing procedure will require precise pixel for pixel location. Any offset could reduce the integrity of the results.
Figure 4-14. The thermal imaging unit focal distance is adjusted to optimize the pixel resolution as well as the optical focus.

Post processing of image data requires pixels to lineup. Three images were acquired for every fracture scenario, both before and after, as well as for each applied current level. A Matlab program was developed which averages each set of three images, so as to mitigate any random anomalies. The baseline image for each fracture condition was subtracted from the image at various current loads (0.1 - 0.5 Amps). The resultant eliminated the consistent environmental disturbances, but could not repair the quality at lower current loads. Using further image analysis features in Matlab, the non-fractured version at each relative current load was subtracted from the fractured version at the corresponding current load. This effectively highlighted locations were temperature increases occurred in the stent, thus highlight regions of fracture.
Figure 4-15. Original thermal image acquired of a double connector fracture, including noise and thermal reflection.

Figure 4-16. Resultant of 12 images processed that highlights temperature distribution in the same double connector fractured stent.

In images above, it may appear obvious where the heat is concentrated. This isn't necessarily the case in every example, moreover the image processing provides a better...
gradient which differentiates features. At this stage, it is not an ideal method for extracting numerical data but yields important information on relative temperature distribution.

### 4.2.3 Discussion

The numerical simulation provided results which seemed intuitive and plausible. Upon conducting the resistance experiment, it was recognized that an objective correlation would be difficult to define. The 100 mA current in the experiment proved insufficient to attain the temperature changes achieved in the numerical simulation. Subsequent increases of 100mA were performed to account for potential error as a result of power factor. However, by observing the experimental results, similar temperature profiles to simulation can be noted. Additionally, the "hot-spots" were located in the specific region of the fracture, supporting the original hypothesis.

<table>
<thead>
<tr>
<th>Current (Amps)</th>
<th>0</th>
<th>0.1</th>
<th>0.2</th>
<th>0.3</th>
<th>0.4</th>
<th>0.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Std. Dev. (°C)</td>
<td>0.42</td>
<td>0.04</td>
<td>0.25</td>
<td>0.38</td>
<td>1.76</td>
<td>0.67</td>
</tr>
</tbody>
</table>

Table 4-4. Standard deviations of baseline samples at various current levels to assess consistency of baseline metric.

Trends in the experimental thermal data appear to mirror that of the resistance data. However, comparing the numerical simulation to the experimental results is more complex. During the low current applications, differentiating thermal effects from the standard deviation of non-fractured stent segment heating is difficult and inconclusive. Standard deviation of non-fractured samples is somewhat inconsistent for a baseline measurement. This negatively affects the sensitivity of results at lower current loading levels. At 400 mA, thermal trends are more perceptible, yet in the case of the connector fracture, not fully understood.
Because joule heating is a function of resistance, any inexplicable trends in resistance would extend to thermal effects as well. This is observed in the experimental results. Additionally, nitinol, is thermally reflective thus being susceptible to the introduction of "thermal noise" from the surrounding environment. Measures were taken to mitigate noise where possible (enclosure) and subtract it when not possible (zero power image subtraction). However, given equipment constraints, this is not a real-time process, therefore only allowing for post-processed images to be analyzed on a relative scale, not absolute.
Figure 4-18. Locations of increased temperature for a long strut fracture (experimental).
Figure 4-19. Locations of increased temperature for a short strut fracture (experimental).
Thermal Signature of Various Fracture Types
Relative Increase with Respect to Unfractured State (Experimental)

Connector Fracture

100 mA

200 mA

300 mA

400 mA

500 mA

Figure 4-20. Locations of increased temperature for a connector fracture (experimental).
Figure 4-21. Locations of increased temperature for a double connector fracture (experimental).
The max/min temperature data included external thermal noise. Because of this effect, it is understandable that lower current settings don't provide the temperature range anticipated. At higher current settings, the material would heat past the thermal noise boundary and become the dominant infrared source. Below this threshold, it is likely drowned out. To improve the signal-to-noise ratio, environmental conditions could be optimized and specific ranges of infrared light could potentially be filtered.
Chapter 5: Conclusions and Future Work

5.1 Summary

In summation, this body of work has made an array contributions, including:

1. Stent fracture modes were identified and classified according to a common standard.
2. Methods for detecting fractures in stents were reviewed and a resistance method of fracture detection was proposed.
3. Methods for locating fractures in stents were reviewed and a novel method of thermal electric identification was proposed.
4. A theoretical model of stents relative change in resistance post fracture was generated.
5. A finite element thermal-electric model of a clinical stent was created, and solved using real material properties, boundary and initial conditions.
6. Various fracture types were simulated in a finite element environment using clinically validated fracture modes
7. Various fracture types were experimentally tested and analyzed using clinically validated fracture modes.
8. Hypothesis that fractures have a detectable affect on stent resistance was supported.
9. Hypothesis that fractures have a detectable affect on a stents thermal signature when joule heating is induced, was supported.
5.2 Conclusions

The results indicate that resistance is significantly affected by the presence of a fracture. The qualitative observations appear to be consistent with the theoretical and numerical models, as well as general expectations. Additionally, the sensitivity analyses are an important measure of feasibility in a 'real-world' environment. It was shown that the "detectability" of such changes is much greater than anticipated, a positive outcome for future implementation. As with the resistive analysis, locating fractures with a thermal imaging routine is shown to be feasible. The hot spots occurred in the regions where fractures were known to exist and maximum temperatures generally exceeded the unfractured maximum temperature. The concepts tested have been validated and provide a benchmark for further development of this approach.

5.3 Future Work

The ability to identify fractures through resistive and thermal approaches has been experimentally validated, however improved correlation between theoretical models, numerical simulation and experimental results is needed. Better understanding of the affects of Nitinol phase change on resistivity could improve correlation. With regard to other bare metal stents such as stainless steel, properties are well established. Moreover, stainless steel has a significantly higher phase change temperature. For standard materials, an analysis could be much more in line with numerical simulation.

Several process improvements may also contribute to higher quality experimental results. If sensor noise and thermal reflection can be mitigated, greater sensitivity can be achieved with lower applied current. It is not only significant from a data acquisition
standpoint, but also reduces the temperature which the nitinol is heated to, thus reducing any phase change effects. Eventually, an automatic experimental setup that would be beneficial to the dynamic observation of stents in fatigue testing cycles. Additionally, process improvements can be investigated to provide more pure data under lower electric loading conditions.

Additional efforts will encompass an alternative method for fracture detection by investigating acoustic detection of fractures in stents. A similar concept to resistive and thermal signatures, fractures release energy in the audible domain. Using passive sensors to record microfractures in a fatigue cycle may provide an alternative method for identifying the structural fatigue and failure of stents.
REFERENCES


Appendix-A

%Fractured Stent Resistance Calculation
%Austen Scudder
%Advanced Manufacturing and Design Lab - Avdeev
%UW - Milwaukee
%First write: 6/23/11

%Edited by:
%Edit date:

clear
clc

% Initial conditions
Sstrut=3; %Relative resistance for short strut (ohms)
Lstrut=5; %Relative resistance for long strut (ohms)

Snum=4; %Number of unfractured short struts
Lnum=4; %Number of unfractured long struts
Rnum=15; %Number of rings in stent

URing=1/(((Snum/Sstrut)+(Lnum/Lstrut))); %Effective resistance of unfractured ring (parallel equivalent)
Ustent=URing*Rnum; %Resistance of unfractured stent

%Case #1: Short strut fracture
S_FRing=1/(((Snum-1)/Sstrut)+(Lnum/Lstrut));
S_Fstent=URing*(Rnum-1)+S_FRing; %Resistance of fractured stent (short strut)

'Short Strut Fracture - Percent Increase in Resistance'
S_DelR=((S_Fstent/Ustent)-1)*100

%Case #2: Long strut fracture
L_FRing=1/(((Snum/Sstrut)+((Lnum-1)/Lstrut)));
L_Fstent=URing*(Rnum-1)+L_FRing; %Resistance of fractured stent (long strut)

'Long Strut Fracture - Percent Increase in Resistance'
L_DelR=((L_Fstent/Ustent)-1)*100

%Case #3: Connector fracture (most likely)
C_FRing=1/(((Snum-1)/Sstrut)+((Lnum-1)/Lstrut));
C_Fstent=URing*(Rnum-1)+C_FRing; %Resistance of fractured stent (connector)

'Connector Fracture - Percent Increase in Resistance'
C_DelR=((C_Fstent/Ustent)-1)*100

%Case #4: Double Connector fracture
C2_FRing=1/(((Snum-2)/Sstrut)+((Lnum-2)/Lstrut));
C2_Fstent=URing*(Rnum-1)+C2_FRing; %Resistance of fractured stent (double connector)

'Connector Fracture - Percent Increase in Resistance'
C_DelR=((C2_Fstent/Ustent)-1)*100
Appendix-B

%Fractured Stent Resistance Calculation
%Austen Scudder
%Advanced Manufacturing and Design Lab - Avdeev
%UW - Milwaukee
%First write: 11/20/11

%Edited by:
%Edited date:
clc
clear

% Program for image analysis of stents

% Import all pictures from testing

% Original Stent (NOT FRACTURED)
 a1N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\a1.png');
 a2N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\a2.png');
 a3N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\a3.png');
 a4N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\a4.png');
 a5N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\a5.png');
 b1N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\b1.png');
 b2N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\b2.png');
 b3N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\b3.png');
 b4N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\b4.png');
 b5N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\b5.png');
 c1N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\c1.png');
 c2N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\c2.png');
 c3N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\c3.png');
 c4N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\c4.png');
 c5N = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\c5.png');

 abaseN = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\abase.png');
 bbaseN = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\bbase.png');
 cbaseN = imread('C:\Users\AKS\Desktop\Experiments\LONG\NF\cbase.png');

% Damaged Stent (FRACTURED)
 a1F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\a1.png');
 a2F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\a2.png');
 a3F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\a3.png');
 a4F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\a4.png');
 a5F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\a5.png');
 b1F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\b1.png');
 b2F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\b2.png');
 b3F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\b3.png');
 b4F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\b4.png');
 b5F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\b5.png');
 c1F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\c1.png');
 c2F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\c2.png');
 c3F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\c3.png');
 c4F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\c4.png');
 c5F = imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\c5.png');
abaseF=imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\abase.png');
bbaseF=imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\bbase.png');
cbaseF=imread('C:\Users\AKS\Desktop\Experiments\LONG\Frac\cbase.png');

%Picture Averaging Operations

%Non-Fractured Averages
baseN=(abaseN+bbaseN+cbaseN)/3;
N1=(a1N+b1N+c1N)/3;
N2=(a2N+b2N+c2N)/3;
N3=(a3N+b3N+c3N)/3;
N4=(a4N+b4N+c4N)/3;
N5=(a5N+b5N+c5N)/3;

%Fractured Averages
baseF=(abaseF+bbaseF+cbaseF)/3;
F1=(a1F+b1F+c1F)/3;
F2=(a2F+b2F+c2F)/3;
F3=(a3F+b3F+c3F)/3;
F4=(a4F+b4F+c4F)/3;
F5=(a5F+b5F+c5F)/3;

%Subtract base (no power) conditions for Delta Temperature

%Non-Fractured Delta Pictures
D1N=N1-baseN;
D2N=N2-baseN;
D3N=N3-baseN;
D4N=N4-baseN;
D5N=N5-baseN;

%Fractured Delta Pictures
D1F=F1-baseF;
D2F=F2-baseF;
D3F=F3-baseF;
D4F=F4-baseF;
D5F=F5-baseF;

%Saving Delta Pictures

%Non-Fractured Saves
imwrite(D1N,'C:\Users\AKS\Desktop\Experiments\LONG\OutputN\D1N.png');
imwrite(D2N,'C:\Users\AKS\Desktop\Experiments\LONG\OutputN\D2N.png');
imwrite(D3N,'C:\Users\AKS\Desktop\Experiments\LONG\OutputN\D3N.png');
imwrite(D4N,'C:\Users\AKS\Desktop\Experiments\LONG\OutputN\D4N.png');
imwrite(D5N,'C:\Users\AKS\Desktop\Experiments\LONG\OutputN\D5N.png');

%Fractured Saves
imwrite(D1F,'C:\Users\AKS\Desktop\Experiments\LONG\OutputF\D1F.png');
imwrite(D2F,'C:\Users\AKS\Desktop\Experiments\LONG\OutputF\D2F.png');
imwrite(D3F,'C:\Users\AKS\Desktop\Experiments\LONG\OutputF\D3F.png');
imwrite(D4F,'C:\Users\AKS\Desktop\Experiments\LONG\OutputF\D4F.png');
imwrite(D5F,'C:\Users\AKS\Desktop\Experiments\LONG\OutputF\D5F.png');
\%Evaluate the difference between Non-Fractured heating and Fractured heating

Dif1=D1F-D1N;
Dif2=D2F-D2N;
Dif3=D3F-D3N;
Dif4=D4F-D4N;
Dif5=D5F-D5N;

\%Saving Overall Difference Pictures

\%Saving Fractured minus Non-Fractured
imwrite(Dif1,'C:\Users\AKS\Desktop\Experiments\LONG\OutputDif\Dif1.png');
imwrite(Dif2,'C:\Users\AKS\Desktop\Experiments\LONG\OutputDif\Dif2.png');
imwrite(Dif3,'C:\Users\AKS\Desktop\Experiments\LONG\OutputDif\Dif3.png');
imwrite(Dif4,'C:\Users\AKS\Desktop\Experiments\LONG\OutputDif\Dif4.png');
imwrite(Dif5,'C:\Users\AKS\Desktop\Experiments\LONG\OutputDif\Dif5.png');
Appendix-C

finish
/clear,nostart

! ************************ thermal boundary conditions and material properties
Sigcurr=.1 !Total Current through system
meshlvl=2 ! Mesh tightness (X by X)
freeh=25E-6 ! Free convection coefficient of nitinol wire (9-25 J/(mm^2*s*K))

On the role of

! thermoelectric heat transfer in the design of SMA actuators: theoretical modeling and experiment:

! 1995. A. Bhattacharyya et al.p.257

icurr=sigcurr/4 ! Current through each top connector
t0=0 ! Temperature (C) (temp used in Incropera and DeWitt (1984) to develop free convection coefficient)
kxx_sma=0.0086 !martensite... 8.6 W/m * deg. C ! Thermal conductivity, W/m-K
rsvx_sma=.0082 !martensite... 8 milli-ohms * mm ! Electrical resistivity, Ohm*m (Nitinol SE508 Tubing/Wire)
ArcR=2.0 ! Radius of curvature of the arm of stent
xpboex=0.6749680599 ! X-coordinate at extended bottom of outer edge
xpto= 0.1005 ! X-coordinate at top of outer edge
xpmo= 0.39305999 ! X-coordinate at middle of outer edge
xpbo = xpboex-0.0045 ! X-coordinate at bottom of outer edge
xpti= 0 ! X-coordinate at top of inner edge
xpmi= 0.29305999  ! X-coordinate at middle of inner edge

xpbi= xpboex-0.1261532-0.0065  ! X-coordinate at bottom of inner edge

xpbiex= xpboex+0.001  ! X-coordinate at extended bottom of inner edge

ypto= -0.145  ! Y-coordinate at top of outer edge

ypmo= -0.925  ! Y-coordinate at middle of outer edge

ypbo= -1.725  ! Y-coordinate at bottom of outer edge

ypti= -0.145  ! Y-coordinate at top of inner edge

ypmi= -0.925  ! Y-coordinate at middle of inner edge

ypbi= -1.825  ! Y-coordinate at bottom of inner edge

connh=0.22  ! Y-coordinate of the top edge of connector

connb=-0.01  ! Y-coordinate of the bottom edge of connector

connx=xpto-0.005  ! X-coordinate of the side of the connector

R1=3.5  ! Radius of inner cylindrical segment

R2=3.6  ! Radius of outer cylindrical segment

filR=0.01  ! Fillet radius 1

filRs=0.005  ! Fillet radius 2

Lcopy=15  ! copy along Z axis

/prep7

MP, EX, 1, 60.0E3  !* MPA or N/square mm

MP,NUXY, 1, 0.3

MP,DENS, 1, 6.45e-6  ! kg/cubic mm

!TB,SMA,1
!TBDATA,1,520,600,300,200,0.07,0 !* SHAPE MEMORY ALLOY
k,1,xpto,ypto,0
k,3,xpmo,ypmo,0
!k,5,xpboex,ypbi,0
!k,6,xpbi,ypbi,0
k,5,xpmi,ypmi,0
k,7,xpti,ypti,0
k,10,0.4,-0.4,0
!k,22,0.30,-1.2,0
LARC,1,3,10,ArcR,
!LARC,2,5,22,ArcR,
!LARC,6,7,22,ArcR,
LARC,5,7,10,ArcR,
L,1,7
L,5,3
Al,1,2,3,4
ARSYM,X,1, , , ,0,0
Nummrg,all
k,12,0,0,0
k,14,0,-0.045,0
LARC,1,2,14
1,1,2
al,3,8,9
LFILLT,7,2,0.01, ,
Al,11,12,13
local,12,0,0,-0.925,0,,

ccsys,12
WPCSYS,-1
arsym,y,1,2,3,0,0
arsym,y,4,5,0,0
!aadd,1,2,3,4
local,14,0,0,-1.85,0,0,0
ccsys,14
WPCSYS,-1
k,32,xpto,ypto,0
k,34,-xpto,ypto,0
k,36,0,0,0
k,38,xpto+0.05,0,0
k,40,-xpto-0.05,0,0
LARC,34,18,40,1.2,
LARC,32,11,38,1.2,
L,32,34
a1,18,23,27,28,29
aadd,1,2,3,4
aadd,5,6,7
AGEN,1,8,0,0,-0.1,1,1
csys,12
WPCSYS,-1
arsym,y,1,2,3,0,0
arsym,y,9,0,0
local,16,1,0,0.25,0,0,90,0,
csys,16
WPCSYS,-1
k,50,0,0,0
k,52,0,0,4
k,54,3.5,250,0
k,56,3.5,290,0
LARC,54,56,50,3.5,
LGEN,2,39, , ,180, , ,0
L,50,52
ADrag,40, , , , ,41
ADrag,39, , , , ,41
VOFFST,9,-4, ,
VOFFST,3,-4, ,
VOFFST,8,-4, ,
Voffst,1,4,,
Voffst,2,4,,
AINV, 5, 5
AINV, 4, 1
k,100,3.5,250,0
k,102,3.5,290,0
LARC,100,102,50,3.5,
LGEN,2,1, , ,180, , ,0
ADrag,1, , , , ,41
ADrag,2, , , , ,41
AINV, 5, 2
AINV, 4, 4
k,120,3.5,250,0
k,122,3.5,290,0
LARC,120,122,50,3.5,
ADrag,1, , , , ,41
| 81 |

| AINV, 1, 3 |

!-----mesh specification------------

| et, 1, 200, 7 |
| esize, 0.1/meshlvl |
| amesh, all |
| eplot |
| et, 2, 226, 110 |
| esize,,meshlvl |
| type, 2 |
| mp,kxx, 1,kxx_sma |
| mp,rsvx, 1,rsvx_sma |

!-------------

| VOFFST, 2, 0.1, , ! extruding surfaces with mesh |
| VOFFST, 6, 0.1, , ! extruding surfaces with mesh |
| VOFFST, 4, 0.1, , ! extruding surfaces with mesh |
| VOFFST, 3, 0.1, , ! extruding surfaces with mesh |
| VOFFST, 52, 0.1, , ! extruding surfaces with mesh |
| alls |
| aclear, all |

!-----copying v-struts to form a ring------

| VGEN, ,2, , ,11.25, , ,1 |
| VGEN, ,3,4, , ,-11.25, , ,1 |
| VGEN,4,3,4, , ,90, , ,0 |
| vGEN, ,5, , ,22.5, , ,1 |
| VGEN,4,5, , ,90, , ,0 |
| VGEN,3,1, , , ,-22.5, , ,0 |
| VGEN,3,2, , ,22.5, , ,0 |
VGEN,2,18, , ,45, , ,0
VGEN,3,19, , ,22.5, , ,0
VGEN,2,21, , ,45, , ,0
VGEN,3,22, , ,22.5, , ,0
VGEN,2,24, , ,45, , ,0
VGEN,3,25, , ,22.5, , ,0
VGEN,2,21, , ,45, , ,0
VGEN,3,28, , ,22.5, , ,0
VGEN,2,30, , ,45, , ,0
VGEN,3,31, , ,22.5, , ,0
VGEN,2,33, , ,45, , ,0
VGEN,3,34, , ,22.5, , ,0

!----copying the whole ring along z axis-----
!Lcopy=15                    ! number of copies of rings
that we want to make
VGEN,Lcopy,all, , ,,-33.75,1.85, ,0   ! make copies of
the ring
modmsh,detach
vdele,all
adele,all
ldele,all
kdele,all
esel,s,type,,1             ! selects 2d elements
edele,all                   ! deletes 2d elements

!****experiment on creating a COMPLETE fracture by deleting
elements**********
edele,49673,49676          ! deleting two rows of element stacks
at the connector
edele,49701,49704
! the fracture is 8 rings down
! following locations of nodes are in csys,0
! location of node 1: 350871
! location of node 2: 350275

****experiment on creating a PARTIAL fracture by deleting elements**********
!
esel,s,elem,,39810,39828,2 ! this is needed if we want to select elements before deleting
!
edele,39810,39828,2 ! deleting two rows of element stacks at the connector

************element deleting experiment ends*************

alls

csys,0

wpcs,0,-1

midconymin=0.920

midconymax=0.930

nsel,s,loc,y,-midconymin,-midconymax ! selects 1st series nodes at and near mid joint

ring=Lcopy-1 ! number of series of nodes is (number of rings-1)

node merge operation begins*********************

!---creating loop to select all series of nodes

*do,i,1,ring,

    nsel,a,loc,y,-(2.774+i*1.85),-(2.776+i*1.85) ! selects nodes at the joints

*enddo

!-----selection of nodes completed

nummrg,node,2.5e-3 ! merge nodes at mid joint location with a higher tolerance
nsel, inve            ! selects the rest of the nodes
nummrg, node, 1e-4    ! merge all other nodes with a
lower tolerance
alls

!!!!!!!! APPLY BOUNDARY CONDITIONS !!!!!!!!!!

!-----VOLTAGE------!
csys, 0
wpcsys, -1

!SET 1  ! select nodes on top
nsel, s, loc, y, -27.895
nsel, r, loc, x, 1.8, 2.1
nsel, r, loc, z, 2.8, 3.1
d, all, volt, 0       ! apply 0 voltage

!SET 2  ! select nodes on top
nsel, s, loc, y, -27.895
nsel, r, loc, x, -2.8, -3.1
nsel, r, loc, z, 1.8, 2.1
d, all, volt, 0       ! apply 0 voltage

!SET 3  ! select nodes on top
nsel, s, loc, y, -27.895
nsel, r, loc, x, -1.8, -2.1
nsel, r, loc, z, -2.8, -3.1
d, all, volt, 0       ! apply 0 voltage

!SET 4  ! select nodes on top
nsel, s, loc, y, -27.895
nsel, r, loc, x, 2.8, 3.1
nsel, r, loc, z, -1.8, -2.1
d, all, volt, 0       ! apply 0 voltage
alls

!-----CURRENT-----!

!!NOTE: Current is applied to 180 nodes in total (4x45), Total Current = 180*icurr
csys,0
wpcs,0,-1

!SET 1  ! select nodes on top
nsel,s,loc,y,-0.14500
nsel,r,loc,x,-3.1,-3.4
nsel,r,loc,z,-1.2,-1.5
*get,i,node,,count  ! count selected nodes
f,all,amps,icurr/i   ! apply current (45 Nodes)
alls

!SET 2  ! select nodes on top
nsel,s,loc,y,-0.14500
nsel,r,loc,x,1.2,1.5
nsel,r,loc,z,-3.1,-3.5
*get,j,node,,count  ! count selected nodes
f,all,amps,icurr/j   ! apply current (45 Nodes)
alls

!SET 3  ! select nodes on top
nsel,s,loc,y,-0.14500
nsel,r,loc,x,-1.2,-1.5
nsel,r,loc,z,3.1,3.4
*get,k,node,,count  ! count selected nodes
f,all,amps,icurr/k   ! apply current (45 Nodes)
alls

!SET 4  ! select nodes on top !
nsel,s,loc,y,-0.14500
nsel,r,loc,x,3.1,3.4
nsel,r,loc,z,1.2,1.5
*get,m,node,,count ! count selected nodes
f,all,amps,icurr/m ! apply current (45 Nodes)
alls
!--------FREE CONVECTION--------!
tunif,t0
alls
nsel,s,ext
sf,all,conv,freeh,t0
alls
finish
/solu
alls
nlgeom,on
nsubst,10,100,5
solve
outres,all
finish
save,LSF_p1,db
!/post1
!set,last
!/view,1,1,1,1
!/ang,1
!plnsol,temp
!finish