In Vitro Measurements of First and Second Tarsometatarsal Joint Stiffness

Andrew R. Fauth\textsuperscript{1,2}, Andrew J. Hamel\textsuperscript{1}, and Neil A. Sharkey\textsuperscript{1,2,3}
The Pennsylvania State University

This study involved a biomechanical examination of the first and second tarsometatarsal (TMT) joints. In the in vitro testing protocol, physiologic joint moments were applied to the first and second TMT joints of 10 cadaver specimens, which had been dissected to yield only the midfoot components comprising the tarsals, metatarsals, and associated ligaments. The isolated joints were placed in a 37 °C water bath and were independently cycled into and out of dorsiflexion, while angular displacement and resultant joint moments were collected. The specimens were sequentially cycled between zero and peak moment levels of 2.5, 5.0, 7.5, and 10 Nm, after which mean moment-angle curves were plotted for each TMT joint at each loading condition. Least-squares curve-fitting procedures, employing a root mean square error threshold of 0.005 Nm, were used to describe the average overall moment vs. angle relationship of each joint. Curves for the first and second TMT joints exhibited similar behavior. The joints displayed reasonably constant stiffness at low angles of dorsiflexion, followed by rapidly increasing stiffness at higher angles of dorsiflexion. These data provide new insight into the behavior of the TMT joints under load and are valuable for use in numerical models of the foot, as well as in the understanding and treatment of certain foot pathologies.

Key Words: Lisfranc, joint moment, midfoot

Introduction

The joints of the midfoot, especially the tarsometatarsal (TMT) joints, play a crucial role in normal foot function during gait. The TMT joints are the articulations between the five metatarsals and the medial, intermediate, and lateral cuneiforms and the cuboid. One of the main functions of the TMT joints, especially the first and second (which together are known as the Lisfranc joint), is to transfer forces...
incurred by the forefoot during the midstance and pushoff phases of gait. Studies of metatarsal loading have repeatedly shown that the first and second metatarsals experience high moments and resultant strains during the forefoot-loading phase of gait (Gross & Bunch, 1989; Hetherington, Chessman, & Steuben, 1992; Hughes, 1985; Sharkey, Ferris, Smith, & Matthews, 1995). As a result, the first and second TMT joints must be relatively stiff and immobile in order to properly transfer forces to the midfoot and maintain the structural integrity of the foot (Bojsen-Moller, 1979).

Anatomical studies of cadaver feet have shown that the shape of the Lisfranc complex determines the transverse arch of the foot. In addition the Lisfranc joint, by virtue of its location, is an integral component of the longitudinal arch. It has even been suggested that this joint acts as the keystone that maintains the integrity and structure of the arch (Burroughs, Reimer, & Fields, 1998; De Palma, Santucci, Sabetta, & Rapa, 1997). Several studies have documented that the longitudinal arch, in concert with the plantar fascia and other soft tissues, functions to absorb and return energy during gait by acting as a spring (Ker, Bennett, Bibby, Kester, & Alexander, 1987; Saltzman & Nawoczenski, 1995) or as a “windlass mechanism,” as proposed by Hicks (1954). Kitaoka and colleagues (Kitaoka, Lundberg, Luo, & An, 1995) reported on a study of midfoot bone motion under physiologic loading using cadaver feet, and noted that the stiffness affecting one joint in the midfoot had a profound influence on the mobility of the surrounding joints. In other words, the mobility of adjacent joints is limited by conditions such as tarsal arthrosis or arthrodesis.

Following this line of reasoning, if the stiffness of the Lisfranc joint is abnormal, the integrity of the medial longitudinal arch may also be compromised and have the potential to precipitate a variety of clinical disorders. Indeed, it has been widely suggested that decreased TMT joint stiffness due to hypermobility of the TMT joint, especially the first, can lead to painful and problematic foot disorders, such as hallux valgus (Faber, Zollinger, Kleinrensink, et al., 2001; Kura, Luo, Kitaoka, Smutz, & An, 2001; Lakin, DeGnore, & Peinkowski, 2001).

To date there are little detailed data regarding the normal biomechanical behavior of the TMT joints, especially quantitative data collected through in vitro means. Data in the literature concerning the loads imposed on the TMT joints during gait are derived from numerical models of the forefoot, which estimate fairly high loads and joint moments of 8–13 Nm at the first TMT joint and 4–9 Nm at the second TMT joint (Stokes, Hutton, & Stott, 1979). Bench-top studies have focused on the material properties of individual ligament complexes at the first and second TMT joints (Hofstede et al., 1999; Kura et al., 2001; Solan, Moorman, Miyamoto, Jasper, & Belkoff, 2001), but none have reported on the overall structural properties or behavior of an uncompromised, intact TMT joint.

Therefore, the primary objective of this study was to determine the structural behavior of the first and second TMT joints. This was accomplished by applying physiologic moments to the first and second metatarsals of nonembalmed cadaver specimens. In practical terms this information will increase our understanding of the functional anatomy of the foot. Furthermore, quantitative data describing TMT joint stiffness, as well as other midfoot structural properties, provide valuable input to computer models of the lower extremity.
Methods

Ten fresh frozen, nonembalmed, nonpaired cadaver specimens were used for this study. Specimens were purchased through medical suppliers, screened for pathology, and shipped frozen to the Center for Locomotion Studies at The Pennsylvania State University. Each specimen was catalogued and stored in a freezer, and then thoroughly thawed just prior to the experiment in order to preserve the integrity of the soft tissue structures. Four right feet and six left feet, with a specimen age range of 22–81 years (mean age of 68 ± 19) were tested.

Each donor midfoot was carefully separated from the hindfoot at the talonavicular and calcaneocuboid joints yielding a distal segment composed of the cuboid, navicular, the three cuneiforms bones, the five metatarsals, and the phalanges. Each specimen was then further dissected to remove the soft tissues and the phalanges, taking care not to disturb the tarsometatarsal joint capsules and supporting ligaments. The resulting midfoot specimens comprised the navicular, cuboid, cuneiforms, and all five metatarsal bones.

To prepare the specimens for testing, we potted each midfoot segment in polymethylmethacrylate (COE Tray Plastic, GC America Inc., Alsip, IL) securing the navicular, cuboid, and cuneiforms just posterior to the TMT joints. Screws were drilled into the tarsal bones for stability. The first and second metatarsal heads were individually potted in small custom-made aluminum fixtures and were secured with screws. Care was taken during the potting of the metatarsal heads to align the small pots with the long axis of the metatarsal, to ensure that the load applied by the servohydraulic testing machine was oriented perpendicular to the long axis of the metatarsal shaft. Metatarsal measurements were made with a digital caliper, including metatarsal length (with and without the fixture), shaft height and width, and metatarsal head height and width.

The potted midfoot specimens were secured in an experimental testing apparatus (Figure 1) consisting of a custom-made aluminum jig mounted within a 37 °C water tank. The tank was secured onto the platform of a mechanical servohydraulic testing machine (MTS Mini Bionix; MTS Systems Corp., Minneapolis, MN). The distal metatarsal fixture, containing either the first or second metatarsal head, was secured to an aluminum shaft, which in turn was connected to the MTS actuator via a custom linkage. The linkage comprised a multiple bearing configuration allowing for 2-D friction-free translation in the sagittal and frontal planes. Mounted in series with the aluminum shaft was a potentiometer (Novotechnic, Southborough, MA) to record the angular displacement of the TMT joint during loading, and a load cell (Omega 100-lb mini tension/compression cell; Omega, Stamford, CT) to monitor forces. Prior to data collection, each specimen was acclimated to the water bath for approximately 45 minutes.

Each specimen underwent mechanical testing using the MTS system software. The actuator of the servohydraulic system applied a dorsiflexing force (F_M) that remained perpendicular to the long axis of the metatarsal bone throughout the loading cycle, at a distance (d) from the estimated TMT joint center. This led to a bending reaction moment (T) at the TMT joint (Figure 2).

As the metatarsal bone was loaded, it moved through an angle (θ) that was measured by the potentiometer. Operating under the assumption that the testing apparatus was rigid and that the dorsiflexion force F_M was always applied perpendicular to the long axis of the metatarsal, the angle θ was used to represent the
angular deviation of the metatarsal. The calculated moment was then used to derive the stiffness in bending measured in Nm/deg at any given joint angle by taking the instantaneous derivative of the applied moment at that angle.

Prior to testing, the failure moments of the first and second TMT joints were unknown. To avoid immediate overload, each specimen was cycled to successively increasing peak moment levels of 2.5, 5.0, 7.5, and 10 Nm. Testing to progressively increasing peak moment levels yielded large amounts of data to describe the behavior of these joints under variable moments. All specimens were conditioned under force control for 10 cycles at each peak moment level. Following the conditioning regimen, we established displacement limits corresponding to the desired peak moment levels and collected six cycles of data under displacement control using a sine wave-loading pattern with a frequency of 0.05 Hz.

Calculated data were processed in Matlab (Mathworks 2000) using a custom routine in which angular displacement and force data were input and compiled for each specimen. All raw data were filtered using a 4th-order, low-pass Butterworth filter with a cutoff frequency of 0.4 Hz. Displacement, load, and rotational data were sampled at a rate of 100 Hz. Joint moment vs. angular displacement data were averaged over the six cycles for each specimen at each testing level, and the joint stiffness (defined as the instantaneous derivative of joint moment) was calculated at 1-deg increments over the entire range of motion.

Bending moment vs. angular displacement data were then averaged across all specimens to obtain mean moment-angle curves for each joint and testing level.
A least-squares-fit routine was employed to fit polynomials, $T(\theta)$, to these relationships. The procedure fit polynomials of successively increasing order to the average curves until a root mean square error of less than 0.005 Nm was reached. The equation of the best-fit polynomial was used to characterize the overall moment vs. angle behavior of the joints.

**Results**

Ten first TMT joints and nine second TMT joints yielded usable data at the 2.5-, 5.0-, and 7.5-Nm peak moment levels. Data from one of the second TMT specimens had to be discarded due to technical error. Only six first and six second TMT joints survived the 10-Nm peak moment test, due to specimen failure at the TMT joint, the metatarsal head pot, or the metatarsal shaft during testing.

Moment vs. angle plots for the first and second TMT joints are shown in Figures 3 and 4, respectively. Both displayed nonlinear behavior regardless of the peak moment to which they were cycled, demonstrating the dependence of joint stiffness on flexion angle. As the peak moment level increased, higher-order polynomials were required to fully characterize the moment vs. angle relationships. Polynomial equations for each curve [$T(\theta)$] are displayed in Table 1.

Overall mean and standard deviations of joint moment and stiffness for the first and second TMT joints during the 5-Nm peak moment tests are listed in Table 2. From the table it can be observed that at low angles of dorsiflexion, the moment
Figure 3 — Mean moment vs. angle plots characterizing first TMT joint behavior when cycled to different peak joint moment levels. Black lines represent mean moment vs. angle curves for cadaver specimens, and shading represents ±1 SD. Dashed lines represent fitted polynomials $T(\theta)$ that were used to characterize mean moment angle curves. $N = 10$ for the 2.5-, 5.0-, and 7.5-Nm plots, and $N = 6$ for the 10-Nm plot.

vs. angular displacement curves for the first TMT joint displayed low stiffness values that remained reasonably constant (less than 0.5 Nm/deg). However, as the angular displacement increased, the joint stiffness changed markedly as a function of dorsiflexion. For example, in the 5-Nm loading trial at 10 deg of dorsiflexion, the mean stiffness increased exponentially to 1.64 Nm/deg.

Moment vs. angular displacement curves for the second TMT joint demonstrated similar behavior. Much like the first TMT, the mean stiffness values measured during the 5-Nm peak moment test at small angular deviations were small and relatively constant (under 0.3 Nm/deg), while at higher angular deviations the mean stiffness rapidly increased as a function of dorsiflexion angle. In this case, at 10 deg of dorsiflexion during the 5-Nm peak moment test, the mean stiffness measured for the second TMT was 1.61 Nm/deg (Table 2).

Discussion

The results from this in vitro study provide unique information regarding the behavior of the first and second TMT joints during metatarsal loading. The literature describing the mechanical behavior of the TMT joints and the supporting liga-
ments is sparse, and there are little if any data describing the structural properties of whole, uncompromised TMT joints. The present experiments yielded moment vs. angle curves for the first and second TMT joints that suggest two different phases of loading. The initial portions of the moment vs. angle plots were characterized by low and reasonably constant stiffness values, which may be indicative of the normal, functional range of loading, when the capsular and ligamentous constraints remain relatively slack. The latter portion of the moment vs. angle plots, characterized by rapidly increasing stiffness, suggest an overloaded region in which the joint is pushed beyond its normal range of motion and the passive constraints act to resist further displacements.

A positive relationship was noted between the order of the polynomial curve fit and the magnitude of the peak moment employed for testing, i.e., progressively higher order polynomials were required to fully characterize moment vs. angle curves generated from tests conducted at progressively greater peak moments. This trend is not surprising, given the viscoelastic nature of the tissues encapsulating each joint, and likely reflects an order effect whereby joint stiffness is dictated to
Table 1  Polynomial Fit Equations for Mean TMT Moment-Angle Plots

<table>
<thead>
<tr>
<th>Load (Nm)</th>
<th>Polynomial fit T (θ)</th>
</tr>
</thead>
<tbody>
<tr>
<td>First TMT Joint</td>
<td></td>
</tr>
<tr>
<td>2.5</td>
<td>(7.9×10^{-3}) θ² − (2.55×10^{-2}) θ + (0.128)</td>
</tr>
<tr>
<td>5.0</td>
<td>(3×10^{-4}) θ³ + (1.55×10^{-3}) θ² − (6.3×10^{-3}) θ + (7.8×10^{-2})</td>
</tr>
<tr>
<td>7.5</td>
<td>(1.0×10^{-4}) θ⁴ − (2.2×10^{-3}) θ³ + (2.39×10^{-2}) θ² − (9.09×10^{-2}) θ + (1.95×10^{-1})</td>
</tr>
<tr>
<td>10.0</td>
<td>(1.0×10^{-4}) θ⁴ − (2.2×10^{-3}) θ³ + (2.39×10^{-2}) θ² − (6.65×10^{-2}) θ + (1.632×10^{-1})</td>
</tr>
<tr>
<td>Second TMT Joint</td>
<td></td>
</tr>
<tr>
<td>2.5</td>
<td>(5.6×10^{-3}) θ² − (2.79×10^{-2}) θ + (0.123)</td>
</tr>
<tr>
<td>5.0</td>
<td>(2×10^{-4}) θ³ + (1.8×10^{-3}) θ² − (1.66×10^{-2}) θ + (1.05×10^{-1})</td>
</tr>
<tr>
<td>7.5</td>
<td>(2.0×10^{-4}) θ⁴ + (3.5×10^{-3}) θ³ − (2.89×10^{-2}) θ² + (1.32×10^{-1}) θ − (2.98×10^{-1})θ +3.21×10^{-1})</td>
</tr>
<tr>
<td>10.0</td>
<td>(2.0×10^{-4}) θ⁴ − (4.6×10^{-3}) θ³ + (4.84×10^{-2}) θ² − (2.08×10^{-1}) θ + (3.58×10^{-1})</td>
</tr>
</tbody>
</table>

Table 2  Mean (±SD) Joint Moments and Stiffness Values Over First 10 deg of Dorsiflexion for TMT Joints During 5-Nm Peak Moment Trial

<table>
<thead>
<tr>
<th>Angle (deg)</th>
<th>First TMT Joint</th>
<th>Second TMT Joint</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Moment (Nm)</td>
<td>Stiffness (Nm/deg)</td>
</tr>
<tr>
<td>1</td>
<td>0.157 ± 0.77</td>
<td>0.118 ± 0.12</td>
</tr>
<tr>
<td>2</td>
<td>0.260 ± 0.13</td>
<td>0.159 ± 0.22</td>
</tr>
<tr>
<td>3</td>
<td>0.391 ± 0.44</td>
<td>0.236 ± 0.31</td>
</tr>
<tr>
<td>4</td>
<td>0.571 ± 0.73</td>
<td>0.334 ± 0.40</td>
</tr>
<tr>
<td>5</td>
<td>0.831 ± 1.10</td>
<td>0.452 ± 0.49</td>
</tr>
<tr>
<td>6</td>
<td>1.203 ± 1.50</td>
<td>0.594 ± 0.58</td>
</tr>
<tr>
<td>7</td>
<td>1.725 ± 1.90</td>
<td>0.790 ± 0.69</td>
</tr>
<tr>
<td>8</td>
<td>2.439 ± 2.40</td>
<td>1.020 ± 0.08</td>
</tr>
<tr>
<td>9</td>
<td>3.389 ± 3.00</td>
<td>1.390 ± 1.20</td>
</tr>
<tr>
<td>10</td>
<td>4.609 ± 3.70</td>
<td>1.640 ± 1.20</td>
</tr>
</tbody>
</table>
some degree by the immediate loading history, in the present case increasingly
greater deformations. Additionally, all cyclic testing was conducted at the same
frequency regardless of peak moment and displacement. Tests conducted at high
peak moments would therefore tend to have slightly greater loading rates, which
may have also influenced the shape of the moment vs. angle curves.

Studies of intact cadaver feet have estimated the functional range of motion
for the first TMT joint to be approximately 2.5 (2–4) degrees of dorsiflexion (Faber
et al., 2001; Ouzounian & Shereff, 1989; Wanivenhaus & Pretterklieber, 1989).
These data suggest that the moments applied to the TMT joints during the present
study may be well beyond the moments normally experienced at these joints. The
presence of a threshold or transition point between the more linear region and the
quadratic region at approximately 3–4 deg of dorsiflexion of the first TMT also
supports the notion that the small moments experienced by the first TMT at angu-
lar displacements below 4 deg are normal, while larger moments beyond the thresh-
old may be abnormal.

Such a behavioral match is less apparent at the second TMT joint. Ouzounian
and Shereff (1989) report the range of motion for this joint to be 1–2 deg of dorsi-
flexion. However, the average transition in these experiments occurred around 4–
5 deg which, although low, is still twice that of previous estimates. If the transitions
measured in the current study were truly representative of the threshold between
normal and abnormal ranges of motion and loading, then we would expect the
second TMT joint to display a lower transition point. On the other hand, the 1–2
deg of motion at the second TMT joint reported by Ouzounian and Shereff (1989)
was measured using cadaver specimens in unloaded, nonphysiological conditions
and may be erroneously low.

The ground reaction forces exerted on the first and second metatarsal heads
during gait produce high moments on the metatarsals. Numerical models and ca-
daver simulations estimate those moments to be between 5 and 13 Nm (Jacob,
2001; Sharkey et al., 1995; Stokes et al., 1979). Data from the current tests suggest
that the moments experienced by the first and second TMT joint may be signifi-
cantly less. According to the moment-angle plots obtained from this experiment
for the first and second TMT joints, a peak joint moment of 8.0 to 13.0 Nm, as
reported by Stokes (1979), would require the angular displacement of the first
metatarsal at the TMT joint to be around 10–12 deg, which is four to five times
greater than the reported ROM at these joints (Faber et al., 2001; Ouzounian &
Shereff, 1989; Wanivenhaus & Pretterklieber, 1989). The present findings suggest
a greater protective role for those extrinsic stabilizing mechanisms that serve to
counteract the dorsiflexion moment at the TMT joint. These mechanisms are both
passive and dynamic; they include the plantar fascia, the deep plantar ligament,
the short and long digital flexors, the quadratus plantae, the adductor hallucis, and
the tibialis anterior.

The present work must be viewed in light of its inherent limitations. Most of
the specimens examined were from older adults with relatively unknown histories,
and younger specimens may behave differently. Furthermore, the small number of
specimens analyzed, combined with the large variability in individual feet, calls
for caution when generalizing these findings to the larger population. Using mean
stiffness values to represent the first or second TMT joint may be appropriate for
computational models, but caution should be exercised due to the large variabil-
ity in stiffness measurements between specimens (Table 2).
Only pure bending moments were applied to the first and second TMT joints in these experiments. No shear or axial loads were imposed on the joints, and only the uncompromised ligaments and residual soft tissue of the TMT joint capsule were tested. It may be that the presence of shear and axial reaction forces at the TMT joints help to stabilize against torsional deformation and alter the joint mechanics and ranges of motion. Indeed, the lack of axial loading may partly account for the discrepancies in joint loading noted between this and earlier work. New experimental techniques will need to be developed and executed to explore this possibility.

It is assumed that the freezing and thawing of the cadaver specimens used in this experiment did not compromise the integrity of the tissues. Furthermore, although much effort was put forth to best represent the in situ conditions of the foot by cyclically loading the samples in a 37 °C water bath, the simulation does not account for the effects of skin and the more superficial tissues. As a result, the observed response of the TMT joint tissues to artificially imposed loads may differ somewhat from the natural response observed in a live person. More important, the viscoelastic behavior of the joint tissues is also influenced by the rate of loading. In this experiment the 0.05 Hz loading rate used was 1/20th of the normal frequency of human walking, which has been estimated to be around 1 Hz (Kadaba, Ramakrishnan, & Wootten, 1990). As a result, the stiffness values observed in these experiments by testing at the slower frequency may not be completely representative of in vivo behavior. However, it is worth noting that previous studies of ligamentous tissue found no appreciable change in stiffness over strain rates ranging up to two orders of magnitude (Peterson, Gomez, & Woo, 1987; Peterson & Woo, 1986). Such data argue that the stiffness values measured in this work may be comparable to those present in vivo.

The results presented herein provide unique insight into the structural behavior of the first and second TMT joints. These data contribute to a basic understanding of the first and second TMT joints and their behavior under load. Furthermore, mathematical descriptions of the stiffness and loading behavior of these TMT joints serve as valuable inputs for numerical models of the foot.

References


Acknowledgments

This work was supported by the National Institute of Health, Grant #5 R01HD37443-02. The authors would like to acknowledge Dr. Peter R. Cavanagh for his insight and contributions to this work, and Dr. Ahmet Erdemir for all of his time and valuable programming assistance.