

# Comparison of lower limb segment forces during running on artificial turf and natural grass

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## Abstract

**Introduction:** Artificial turf, soon after being introduced in the 1980s, became associated with an increased injury incidence in football players. While more recent generations of artificial turf have mitigated the problem, perception of the material is still widely negative. So, the decision to play the 2015 Fédération Internationale de Football Association Women's World Cup in Canada on artificial turf was met with vocal criticism by many players. One common approach is to assess injury incidence to quantify risk differences in playing surfaces. This, however, does not account for possible confounding variables or chronic injuries. Direct measurement of ground reaction forces is difficult because conventional multicamera-based motion capture and force plate equipment are limited in its use outside of dedicated laboratories.

**Methods:** We describe a method of generating realistic force data by using miniature load cells that are installed directly into the weight-bearing structure of the body.

**Results:** Pilot data show a significant ( $p < 0.01$ ) difference in peak forces on artificial turf (272% of body weight) and natural grass (229% of body weight).

**Discussion:** Invasive surgical procedures were avoided by installing the load cell into the prosthesis of an athlete with lower limb loss. As modern prosthetic devices allow a close approximation of able-bodied kinematics and kinetics, such prosthesis-based data are transferable to a general population.

## Keywords

Injury, gait analysis, soccer, football, artificial turf, mobile data collection, load cell, sensors/sensor applications, surfaces, limb prosthetics

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## Introduction

Football, the world's most popular sport,<sup>1</sup> has always been played on natural grass (NG) until the advent of artificial turf (AT) in the 1980s provided a more durable and economical option with lower maintenance costs. Despite technological advancements in AT, many players still perceive it as “too hard”<sup>2</sup> and associate it, among other things, with poorer ball control, greater physical effort, fewer slide tackles, and an increase in injury rate when playing on AT instead of NG.<sup>3</sup> Notably, a number of professional players sued the organisers of the 2015 FIFA Women's World Cup on the precedence of sexism,<sup>4</sup> because the tournament was (unlike Men's World Cup tournaments) held on all AT fields.

Research comparing AT with NG mostly investigated injury risks, generally finding no significant

difference in injury incidence,<sup>5</sup> severity, nature or cause.<sup>6</sup> However, assessing injury incidence to quantify risk differences in playing surfaces does not account for possible confounding variables (practice vs. competition) or cumulative injury mechanisms. Hard surfaces are associated with players' higher energy expenditure, making them more susceptible to injury over time,<sup>7</sup> increasing the chance of developing injuries such as

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medial tibial stress syndrome, chronic ankle sprains, and cartilage degeneration.<sup>8</sup> Attempts to quantify pitch hardness by Clegg-Hammer or Penetrometer had inconclusive results.<sup>9–14</sup> The standardized method used to quantify hardness of North American Football fields (ASTM 355-A) may be limited in replicating realistic impacts of body parts and drawing the appropriate conclusions on injury risk.<sup>15</sup>

Some researchers have investigated ground contact times, however, without finding significant differences between NG and AT ( $p=0.465$ ).<sup>16</sup> Collecting and analysing more sophisticated biomechanics data<sup>17</sup> are problematic outside of a gait laboratory, as most equipment for accurate assessment of kinematics and kinetics by way of inverse kinematics is not easily portable and has a small capture volume. Force plates may also interfere with surface characteristics of the actual field.<sup>18</sup> Wearable data collection equipment can address some of these shortcomings, but is often cumbersome and inaccurate as it has a tendency to shift during movement and produce soft tissue artefacts.<sup>19–21</sup> Soft tissue artefacts associated with skin mounted reflective markers can have magnitudes of up to 40 mm.<sup>22</sup> Wearable sensors may also interfere with the normal motion patterns to be investigated, for instance when they or their respective fixation straps restrict the muscle play or the range of motion of an adjacent joint.<sup>23–26</sup>

Direct measurements of segment forces, unaffected by motion artefacts and error propagation, require sensors placed in line with the load-bearing structure of the body. This has been done before, although not in an athletics context. Researchers have implanted load cells in patients with artificial hip joints to collect most accurate hip joint force data.<sup>27</sup> A similar approach was used to obtain in vivo strain measurements from the tibia bone.<sup>28</sup> A far less invasive way of obtaining data of a comparable quality is presented in prosthesis-integrated load cells in persons with limb loss.<sup>29</sup> Unlike wearable sensors that are attached to the surface of the body, the load cell is part of the structure of the body and thus not subject to motion artefacts. Although prosthesis users are known to display gait asymmetries to compensate for the unilateral loss of structure and function, the current state of prosthetics technology is advanced enough to facilitate kinematics and kinetics similar to those of able-bodied controls, which suggests the translatability of some prosthesis-based data to a general population.<sup>30–34</sup> Prosthetic feet, for instance, is designed to mimic a physically sound foot's shock attenuation, propulsion, and ground reaction lines.<sup>31</sup>

The aim of the here described single-subject study was to demonstrate the utility of load cell-based force measurements to obtain ground reaction forces during running on AT and NG, as a first step to possibly adopting the methodology for future investigations of

the more complex biomechanics of football play on different surfaces. Initial data were collected to address the hypothesis that peak ground reaction forces and force gradients are significantly higher on AT than on NG.

## Methods

One recreational athlete with unilateral trans-tibial limb loss was recruited for participation in this pilot study. Inclusion criteria were a well-fitting endo-skeletal prosthesis and clearance for athletic activity by both a physician and a prosthetist. The University of Pittsburgh International Review Board (IRB) approved all study procedures and written consent was obtained prior to testing.

Kicking kinematics of our participant were compared to those of a sample of professional and recreational football players by qualitative video analysis to confirm that his motion patterns (i.e. knee and hip angles) fell within the respective wide range that was observed in able-bodied football players in this representative activity.<sup>35</sup>

Running trials were conducted at self-selected running speeds with the intention to apply a mathematical correction to the raw data of ground reaction force if necessary to account for possible differences in self-selected running velocity ( $dv$ ) between trials

$$dF = dv \cdot 0.598$$

The change in peak force  $dF$  is here represented as % body weight (BW). This equation is a simplification of a previously published linear function equation that describes the relationship between absolute running velocity and absolute ground reaction force.<sup>36</sup>

A strain-gage based load cell (iPecs Lab, RTC Electronics, Dexter, MI) was installed in the endoskeleton of the limb prosthesis by a credentialed prosthetist, using the standard adapter system that allows easy compatibility of various components within the endoskeleton of a limb prosthesis (Figure 1).<sup>37</sup> The prosthetic alignment that had been previously optimized for the patient was maintained, yet the load cell added approximately 230 g of weight to the device.

Ground reaction force data were collected at 100 samples per second, streamed wirelessly to a laptop computer, and plotted in a fashion similar to laboratory-based gait analysis (i.e. generating a graph that shows the bodyweight normalized ground reaction force over time). Upon acclimatising, which entailed a few steps with the modified prosthesis to get used to the slightly changed weight distribution, and going through his warm-up routine, which entailed stretching and a lap of slow jogging, the participant completed 10 timed runs of about 9.1 m (10 yards) at his self-selected



**Figure 1.** Illustration of the load cell installed in the prosthesis.

running speed; 5 of them on AT and 5 on NG. Rest was allowed after each run. Two-dimensional video data were recorded to allow post hoc velocity calculations. Data were collected at the training facilities of a professional American Football team where practice fields with both playing surfaces were available. Field markings painted on the surface were used as orientation landmarks, resulting in distances being measured in yards rather than metres.

The intermediary step from each running trial on AT was selected for comparison against the respective steps on NG. Data were not smoothed or otherwise processed apart from baseline corrections when necessary. The dependent variables peak forces and force gradients were then compared across surfaces (independent variable) by *t*-test. An alpha of 0.05 was defined as significance criterion prior to analysis. Other independent variables were the subject's body weight and self-selected running speed.

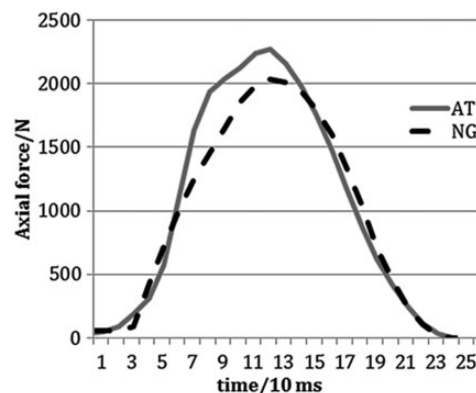
## Results

The participant was 26 years old (173 cm, 88.5 kg) and had a PLUS-M<sup>38</sup> mobility score in the 98% percentile.

Data analysis showed higher peak ground reaction forces ( $p < 0.01$ ) with an average force change equivalent to 43% of the participant's body weight (BW) on AT (272.7% BW, Standard Deviation = 12.3% BW)

**Table 1.** Peak axial shank forces during running on different surfaces (% bodyweight).

	Run 1	Run 2	Run 3	Run 4	Run 5
AT	264.3	268.8	261.2	291.8	277.4
NG	234.7	223.6	230.8	245.2	213.5



**Figure 2.** Example of NG and AT peak forces during running.

when compared to NG (229.6% BW, SD = 11.9% BW), (Table 1). AT caused a steeper initial force gradient than NG before the peak but after the peak gradients were similar for both surfaces (Figure 2).

## Discussion

Results from this pilot data collection, though limited in generalizability, provide some support for the hypothesis that ground contact peak forces are higher on AT than NG. This somewhat contradicts previously published findings that injury rates were not affected by playing surface. One possible explanation is that the here found effect sizes are clinically not significant (i.e. that increasing the ground reaction force by about 40% of the body weight does not have any noticeable adverse effects on the body). However, the accumulated effect of unnecessarily high segment forces from multiple cycles may reach a clinically significant threshold eventually, leading to overuse injuries.<sup>39</sup> In players who are active on different surfaces, as is often the case in football players who may have their games on NG but practice sessions on AT, the need to adapt to different surface hardness may cause an acute injury on either playing surface. The incidence of the injury on NG or AT would follow a somewhat random pattern, which would be consistent with previous findings that injury rates are similar on both turfs.

The utilized methodology is not without limitations. It has been shown that axial segment forces are not entirely comparable to conventionally obtained ground reaction forces.<sup>29</sup> While this possibly affects the

external validity of our findings, the within-participant design assured internal validity. The likely differences in run and sprint kinematics between people with leg prostheses and able-bodied athletes (e.g. the risk that the prosthetic limb is habitually subjected to lower loads than the sound limb) are likewise accounted for by the repeated measures design of this study. Any effect of the playing surface on segment forces would unlikely only occur in players with prostheses, or in players with particular motion patterns. The propagation of ground reaction forces through the body follows essentially the same principles, with only minor deviations being attributable to the presence of an artificial limb. However, given the higher capabilities of the physiological spring-damper apparatus within the biological lower limb, it is possible that effect sizes are exaggerated in participants with a prosthetic leg.

While increasing the sample size in subsequent studies may be challenging (only a small population meets the strict inclusion criteria for the protocol), the principle of direct measurement of segment forces by load cell may be expanded to include shear forces and longer term assessments, and thus help address many of the noted limitations in researching the effect of playing surfaces on injury risk.<sup>9</sup> Combining the load cell with established methods and equipment would allow for more accurate accounting of independent variables, such as the running velocity, and thus mitigate the limitations of this pilot data collection protocol. The described method may therefore be a viable option for studies on the biomechanical effects of turfs.

#### Authors' note

Shea McMurtry is also affiliated with School of Biological Sciences, Georgia Institute of Technology, Atlanta, GA.

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SM: Literature review, protocol design, IRB application, data collection, analysis, interpretation, manuscript writeup.

GF: Study concept, recruitment, assisting in data collection and interpretation, manuscript revisions.

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