Lower Leg Fracture Risk During Football Impacts Using Male, Female, and Child Human Body Models

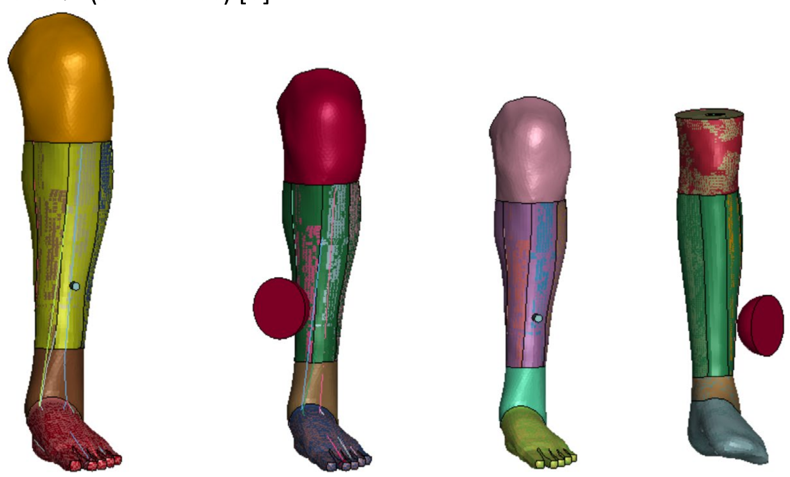
Erin M. Gray, Aryen Shakib, Cheryl E. Quenneville

I. INTRODUCTION

Football (soccer) is the fastest growing and largest participatory sport in Canada, with nearly one million active players joining 270 million worldwide [1]. In-game lower extremity fractures can occur through contact during slide tackles, collisions with goal posts, or with other players. On average, a professional football team will experience one to two fractures per season, with recovery periods of about 2–9 months and increased risk of subsequent injury for the athlete [2]. Wearing shin-guards was made compulsory in 1990 by Fédération Internationale de Football Association (FIFA) to reduce injuries such as contusions, abrasions, lacerations and fractures [3]. Performance standards exist for shin-guards and vary in their profile shape (e.g. spherical and stud) [4-5]. The applied energies are very low (<15 J), not challenging the guards to the level that causes fracture. With increased participation in the sport from women and children (24% global increase in female athletes since 2019), new test standards are warranted that protect against high-energy injuries, with consideration for differences in tolerance due to anatomical differences in athlete sex and size. The purpose of the present study was to identify the conditions of highest risk for these athletes, and the corresponding injury limits to which shin-guards must protect.

II. METHODS

Total Human Model for Safety (THUMS) human body models (HBMs) were used, representing a 50th percentile male (AM50), 5th (AF05) and 50th percentile females (AF50), and a 10-year-old child (10YO) (Fig. 1). The left leg of these models was segmented from the whole body to reduce computational time. The leg was translationally constrained with boundary conditions at the proximally segmented end above the knee and the plantar surface of the foot. A 5 kg stud impactor (10 mm diameter, 16 mm height, European Standard EN 13061) and a 5 kg hemispherical impactor (80 mm diameter, NOCSAE standard) were each positioned at 35% of the tibia height, previously identified as having the highest fracture risk [6]. Impacts were applied using both impactors in three directions (anterior (A), anterolateral (AL), anteromedial (AM)), oriented by rotating by 45° about the centre node of the leg, and constrained to that direction for the full simulation. Fracture response was modelled using the unedited element erosion criterion, which deleted elements from the model once strain exceeded critical values of 2.14% (cortical) and 6.00% (cancellous) [6].



(a) (b) (c) (d)

Fig. 1. Impact configurations for (a) AM50 with stud impactor in A orientation, (b) AF50 with hemispherical impactor in AM orientation, (c) AF05 with stud impactor in A orientation, (d) 10YO with hemispherical impactor in AL direction.

E. M. Gray is an MASc student in the School of Biomedical Engineering at McMaster University in Hamilton, Ontario, Canada. A. Shakib is a PhD candidate in the School of Biomedical Engineering at McMaster University in Hamilton, Ontario, Canada. C. E. Quenneville (quennev@mcmaster.ca) is a Professor in the School of Biomedical Engineering and Department of Mechanical Engineering at McMaster University in Hamilton, Ontario, Canada.

Initial impacts were performed at 90 J (6.00 m/s) and adjusted up or down in increments of 20 J until no fractures were observed (identifying the maximum endurable condition). If fractures were still present at 50 J, increments were reduced by 10 J. The commercial finite element (FE) solver LS-DYNA R12.2.1 was used to conduct all simulations, with a 45 ms termination time to fully observe rebound. The contact force data between the limb and the impactor were collected at 10 kHz. Peak force was recorded for each simulation, as well as the force at fracture (when occurred). For each model, the lowest energy to cause a fracture was identified.

III. INITIAL FINDINGS

All simulations terminated successfully. A total of 90 simulations were run across the four models, with 48 resulting in a fracture. Most fractures occurred in the tibia, with fibular fractures being noted under anterolateral impacts using the stud impactor. The minimum energy to fracture for the AM50 model was 90 J, which occurred in the anteromedial direction using both impactors (Fig. 2). The minimum energy that caused fracture was at 50 J for the AF50 model and occurred at both anteromedial (both impactors) and anterolateral (stud) orientations. For the AF05 model, 40 J led to fractures under the same conditions as the AF50. Finally, the 10YO also experienced fractures starting at 40 J, and these occurred using both impactors in the anterolateral direction.

Sex strongly affected the force to fracture in the models, where the average male model fractured at a mean force of 1925 N and the average female model fractured at 1643 N. Size also affected tolerance, where the 5th percentile female had a mean fracture force of 1460 N. The lowest fracture force was in the 10YO (1255 N). The hemispherical impactor generally caused greater impact forces than the stud but generated fewer fractures.

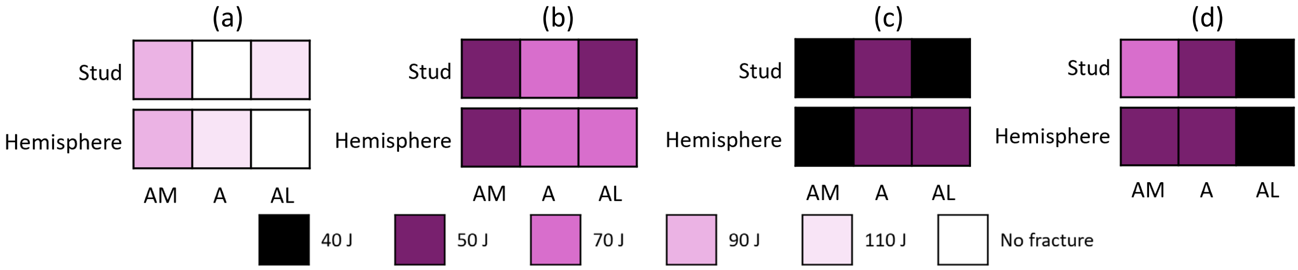


Fig. 2. Injury Matrices for impacts to (a) AM50, (b) AF50, (c) AF05, (d) 10YO leg models. Cell shading indicates the minimum impact energy at which fracture occurs (40–110 J), with white cells representing no fracture.

IV. DISCUSSION

Current shin-guard test standards generally consist of singular metrics (pass/fail) that don't consider the effect of athlete sex or size. The present study highlights the importance of this: the child model fractured at 65% of the force for an average male adult; an average female fractured at 85% of the male limit. This could be attributed to differences in bone mass, cross-sectional area and tibial length, but provides force values to which shin-guards for each demographic should limit. The minimum energy to fracture similarly varied by model, but all greatly exceeded the 15 J currently used in test standards. Given that fractures persist in this sport, it is likely that guards are not being adequately challenged during evaluation.

The most susceptible orientation for the adult models was the anteromedial direction, but the child model was most susceptible to fracture in the anterolateral direction. This may be an artifact of the child model version (4.1 vs 7.1 for the adult models) and is correspondingly more simplified (fewer elements, lack of internal marrow). However, this should be investigated further, and shin-guards should ensure they provide protection in these vulnerable orientations. The effect of impactor shape was likely due to the difference in load-bearing area; as such, any future shin-guard standards should use the threshold corresponding to the shape of the impactor.

These findings demonstrated that the use of an adult male leg tolerance in shin-guard test standards does not adequately evaluate the injury risk for all athletes. Going forward, these results will aid in the development of a new shin-guard standard that considers fracture risk for a wider range of athletes.

V. REFERENCES

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