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Muscle tension increases impact force but decreases energy absorption and pain during visco-elastic impacts to human thighs

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ABSTRACT

Despite uncertainty of its exact role, muscle tension has shown an ability to alter human biomechanical response and may have the ability to reduce impact injury severity. The aim of this study was to examine the effects of muscle tension on human impact response in terms of force and energy absorbed and the subjects' perceptions of pain. Seven male martial artists had a 3.9 kg medicine ball dropped vertically from seven different heights, 1.0–1.6 m in equal increments, onto their right thigh. Subjects were instructed to either relax or tense the quadriceps via knee extension (≥60% MVC) prior to each impact. F-scan pressure insoles sampling at 500 Hz recorded impact force and video was recorded at 1000 Hz to determine energy loss from the medicine ball during impact. Across all impacts force was 11% higher, energy absorption was 15% lower and time to peak force was 11% lower whilst perceived impact intensity was significantly lower when tensed. Whether muscle is tensed or not had a significant and meaningful effect on perceived discomfort. However, it did not relate to impact force between conditions and so tensing may alter localised injury risk during human on human type impacts.

1. Introduction

In sport, during crashes and accidental collisions if there is forewarning it is the norm to 'tense up' to receive an impact, and this is beneficial at the whole body level. Within sports with impacts players tense up both to deliver and receive impacts, and do so differently in varying conditions (Cazzola et al., 2015; Seminati et al., 2016), as well as specifically training and pre-activating muscle groups to reduce impulsive head loading (Eckner et al., 2014). Tensing muscle can reduce limb and torso acceleration and displacement (Muggenthaler et al., 2008), reduce whiplash (Bauer et al., 2001; Brolin et al., 2005), and reduce chest compression (Kemper et al. 2014). Research into the effects of muscle tension on injury risk has been primarily on thoracic stiffness during in vivo crash test studies. Early work found thoracic stiffness increases of 121 – 337% between tensed and relaxed human volunteers (Lobdell et al., 1973; Stalnaker et al., 1973; Patrick 1981; Backaitis and St. Laurent, 1986). However, Kent et al. (2003, 2006) suggested that muscle tension was an unnecessary consideration for crash injury analysis as it made a negligible difference in thoracic stiffness above the injury compression threshold for irreversible injury.

With the development of whole body models for crash test simulations, both finite element models and multibody models (e.g. THUMS, HUMOS, MADYMO), and the development of more biofidelic testing methods for personal protective equipment (Pain et al., 2008; Hrysomallis, 2009; Payne et al., 2016), the effects of muscle tension on impact response needs better quantification. Tensing muscle has been found to change the kinetics of impacts without altering the gross kinematics due to soft tissue intrasegmental motion. Increased muscle tension led to decreased intersegmental tissue movement, increased peak force, decreased time to peak force (Pain and Challis, 2001, 2002, 2004, 2006) and soft tissues were found to account for up to 70% of energy dissipated (Pain and Challis, 2002). Given the ubiquity of tensing musculature for an impact in sport it is surprising there is almost no literature examining effects of muscle tension on localised response to an impact, and whether, or how, it can reduce the risk of localised injuries.

Skeletal muscle contusions are one of the most common muscle injuries (Whiting and Zernicke, 2008; de Souza and Gottfried, 2013). In their review 'Muscle Injury: Review of experimental models' de Souza and Gottfried (2013) described the crush model and the more common, and ecologically valid, blunt non-

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penetrating impacts model, invariably performed on animals. Crisco et al. (1996) performed blunt impacts with a solid nylon impactor onto the gastrocnemius of rats whilst the muscle was tensed for one limb and relaxed for the other. They had hypothesised that impact severity and severity of contusion would be greater in the tensed state but found the peak force decreased and less energy was absorbed during the impact when the muscle was tensed. They ascribed this reversal of their hypothesis to the tensed muscle stopping the impact being dominated by the bone of the limb, with the stiffer muscle reducing the compression of the muscle tissue between the impactor and the bone.

Very few equivalent impactor studies on humans have been carried out even at lower relative impact energies. In the PhD of Hrysomallis (1996) impacts to the relaxed and tensed thigh of human volunteers were reported, but in a published paper of this work, Hrysomallis (2009), only the relaxed impact results were presented. Hrysomallis (1996, 2009) performed a series of drop tests onto the thighs of 18 volunteers using a steel 2.23 kg impactor from between 10 cm and 130 cm, with >90% of drops from 50 cm or less, giving energies up to 21 J. Some of the subjects withdrew after impacts from the higher heights, with more withdrawing after relaxed impacts than tensed impacts. Iwamoto et al. (2015) used a material testing indentor to examine the force deformation relationship of a single subject's upper arm, biceps side, in tensed and relaxed states. They found a marked change in the force displacement curves after 5 mm of depression, with the tensed condition reaching twice the force of the relaxed condition at ~ 13 mm. Muggenthaler et al. (2008) used a 0.93 kg aluminium impactor with impact energies up to 3.64 J to impact the biceps side of the upper arm in both relaxed and tensed conditions for seven subjects. They found slightly higher decelerations and much greater energy absorption in the relaxed condition, similar to Crisco et al. (1996). However, the use of metal impactors can severely limit the biofidelity of testing for human on human impacts as these are between two non-linear visco-elastic bodies (Pain et al. 2008; Tsui and Pain, 2012). The impact velocities and energies are often greatly reduced in metal impactor and anvil tests from those seen in human on human impacts and a more compliant impactor is therefore preferable (Milburn et al., 2001; Pain et al., 2008).

Despite uncertainty of its exact role, muscle tension has shown an ability to alter human biomechanical response and may have the ability to reduce impact injuries or injury severity. The aim of this study was to examine the effects of muscle tension on human impact response in terms of force and energy absorbed and if muscle tension influences the subject's perception of pain. A visco-elastic impactor was used, allowing impacts with greater energies than previously utilised and a more biofidelic human on human type impact. It is hypothesised that being tensed will reduce the severity of the impact as determined by mechanical and subjective measures.

2. Methods

Seven male martial artists (age 27 ± 8 yrs, height 1.76 ± 0.04 m, mass 83.2 ± 8.0 kg) provided informed consent to participate in this study in accordance with the protocol outlined by the Loughborough University Ethical Advisory Committee. Subjects were selected based on their familiarity with, and tolerance to, impacts to the legs, with each subject training for at least four hours per week in martial arts. This ensured that the impact intensity levels were not unfamiliar and that they would be able to distinguish between perceived intensities. Subject numbers were determined from a power analysis for an ANOVA to detect force changes, with a power of 0.8 and p < .05, based on the force results from Crisco et al. (1996) and kept to the minimum subject number needed for ethical reasons. The thigh was chosen as the impact area as it is a favoured target in full contact martial arts with good impacts resulting in the common injury of a dead leg (muscle contusion and numbness) (Bracker, 2012). A dead leg is also a common injury in other impact sports such as rugby (Micheli, 2010). The thigh is a relatively safe place to test given its large ratio of muscle to bone mass and that the major blood vessels and largest nerves are more posteriorly positioned in the thigh. Thus it is likely that discomfort will be due to muscle compression and if there was to be an unexpected injury it would likely only be a muscle contusion.

Each subject sat in an upright posture with their right thigh resting on the middle of a 0.5 m wide bench with the right foot planted flat on the ground. Bench height was adjusted until a resting knee angle of 90° was achieved. The left leg was positioned off the side of the bench, just posterior and inferior to the right leg, to provide an unencumbered view of the impacted thigh. After a short warm-up, consisting of sub-maximal isometric knee extensions, each athlete performed a single isometric maximum voluntary contraction (MVC) for knee extension against an ankle strap placed 2 cm proximal to the medial malleolus, and positioned perpendicular to tibial motion during knee extension/flexion. The strap connect via a steel cable to a bolt screwed into a Kistler 9281b force plate (Winterthur, Switzerland).

In each impact trial, a 3.9 kg medicine ball was dropped vertically from one of seven different heights, 1.0–1.6 m in increments of 0.1 m, onto the right thigh. Subjects were instructed to either relax (no muscle contraction) or tense the quadriceps via knee extension (≥60% MVC) prior to each impact. Muscle activation state was monitored via the force plate readings and visually in real time to check there was no contraction, and visually in the slow motion video between drops. Immediately post-impact, participants were asked to rate their level of discomfort ranging from 0 ('No Pain') to 10 ('Extreme Pain') on a Borg CR10 pain scale. In total each subject was exposed to no more than 20 impacts to obtain a good impact at each height, spread over two sessions with more than 2 days between sessions. This helped to reduce muscle fatigue and bias of their perceived discomfort due to the area becoming overly sensitized from repeated impacts. Drop heights were randomized, but care was taken to ensure that higher energy impacts were not administered in succession to allow for a period of recovery. Subjects were blinded both visually and aurally to the drop heights and the exact time of impact, and time between impacts was at least 120 s.

Two F-scan pressure insoles (Tekscan, Boston, MA) were wrapped around the quadriceps of the right thigh for each subject and secured with electrical tape. Sensors were orientated to maximize the measuring area but limit the overlap between sensors, and to prevent creasing or folding of the sensors (Fig. 1). In addi-

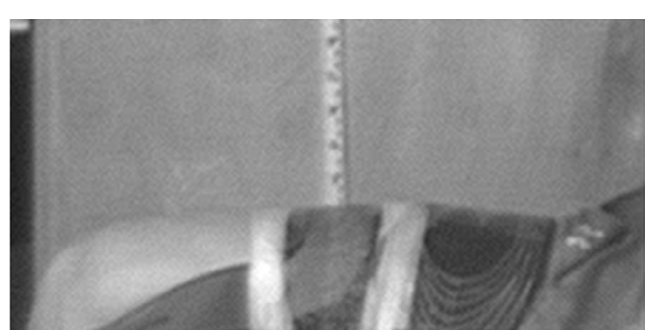

Fig. 1. Two F-scan sensors taped over the top and sides of the thigh to minimise overlap and to avoid creasing or folding.

tion to the static calibration specified by the manufacturer sensors were dynamically calibrated using the technique outlined in Pain et al. (2008). Tekscan sensors sampled at 500 Hz and were used to calculate impact force, contact time and time to peak force.

A Phantom V4 High Speed Camera (512×512 pixels, Vision Research, Wayne, NJ), sampling at 1000 Hz, recorded each ball-thigh impact in a calibrated area of 0.6×0.6 m. In each trial, the three frames pre- and post-impact were digitised three times at sub pixel resolution using SimiMotion software (Unterschleissheim, Germany). In each frame, the top-, bottom-, left- and right-most points of the ball were digitised. These points were averaged to determine the ball centre and changes in its position were used to calculate inbound and outbound velocity of the impactor in the vertical and anterior-posterior directions. As only one camera was used, any impacts with medial-lateral motions were discarded from analysis and these were identified in a two step process. In order to stop a second impact onto the thigh the medicine ball was caught by a researcher stationed in front of the thigh and if they considered that they caught the ball to their right it was an impact with medial-lateral motion. The second part was all impact videos were visually checked whilst saving them for medial-lateral impacts using the leg and the scale marked in line with the drop of the ball.

To determine the energy lost in deforming the medicine ball for each trial, a function was derived for energy lost at a given impact force for the medicine ball undergoing an impact against a rigid object. To obtain this function, the medicine ball was dropped vertically, between heights of 0.50–1.60 m, onto the force plate whilst recorded by the Phantom video camera, sample rates 2000 Hz and 1000 Hz respectively. Energy loss in deforming the ball, $E_{balldeformation}$, was calculated using the difference between drop and rebound velocities, as described in the previous paragraph. This accounted for all energy losses during the ball force plate impact including any minimal frictional losses. Energy lost was plotted against impact force and a power function was fitted to predict energy lost at other impact forces (Fig. 2).

Energy absorbed by the thigh during impact was then calculated using Eq. (1):

$$E_{absorbed} = \underbrace{\frac{1}{2} m v_{impact}^2}_{\text{impact energy}} - \underbrace{\frac{1}{2} m v_{rebound}^2}_{\text{rebound energy}} - E_{ball\,deformation} \quad (1)$$

Two way ANOVAs for repeated measures with a Tukey-Kramer pairwise comparison were performed for muscle tension and drop heights using Matlab (The MathWorks Inc., Natick, MA, USA). Contact time and time to peak force were compared for the grouped drop heights between conditions with a paired *t*-test. Kruskal-Wallis was used for the ordinal discomfort measures. Significance was set at $p < .05$.

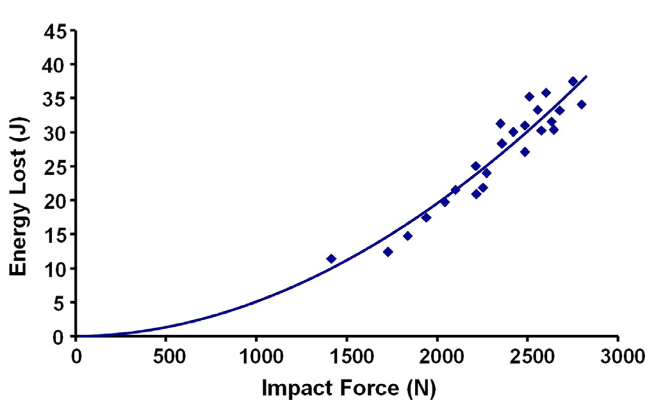

Fig. 2. Energy lost deforming the medicine ball plotted against impact force measured by the force plate. This represents the total energy lost by any means by the ball during the impact.

3. Results

Mean activation was 65% of MVC contraction. The energy lost when deforming the medicine ball for a given impact force was well described ($R^2 = 0.875$) by the power function:

$$Energy = 8 \times 10^{-6} \cdot Force^{1.94} \quad (2)$$

Impacts onto the thigh of up to 61 J were achieved with 50–61% of the impact energy absorbed by the thigh in the tensed condition and 63–70% in the relaxed condition (Table 1), representing higher energies than previous studies and a more human on human scenario. Tensed peak impact forces were on average 11% higher than in the relaxed condition, and energy absorbed was 15% lower in the tensed condition than the relaxed condition (Table 1). Across all heights peak impact force was significantly higher in the tense condition than the relaxed condition ($F = 32.616$, $p = .029$), and energy absorbed was lower in the tensed condition than the relaxed condition ($F = 28.62$, $p = .033$). There was no significant change in impact force with drop height ($F = 2.422$, $p = .26$) but there was a significant change in energy absorbed with drop height ($F = 5.372$, $p = .007$). Perceived impact intensities for tensed muscle, 1.4 ± 0.8, were significantly lower than that for relaxed muscle, 4.2 ± 1.1 (p .016) (Table 2). Tensed impacts had significantly lower time to peak force compared to relaxed impacts ($p = .003$) whilst contact times were not significantly different ($p = .38$) (Table 2). High-speed video revealed qualitatively greater thigh deformations in the relaxed condition, whilst impactor deformation was greater in the tensed condition (Fig. 3), but it was not possible to reliably measure this across subjects with the single camera.

4. Discussion

This study aimed to examine the effects of muscle tension on human impact response during a visco-elastic impact in terms of force and energy absorbed and if muscle tension influences the subject's perception of pain. The hypothesis that being tensed will reduce the severity of the impact measures was supported for the subjective perception measures with significantly lower pain index scores when tensed. With regard to significant mechanical measures, across all impacts force was 11% higher, energy absorption was 15% lower and time to peak force was 11% lower when tensed. The results seen here exhibited the same trend as Hrysomallis (1996, 2009) which attributed an increased force and decreased time to peak force to a greater hardness or resistance to deformation. The results found in this study are in agreement with those of Hrysomallis (1996, 2009) who also observed a lower level of discomfort when tensed. For impacts to a well-muscled area higher impact forces, from higher peak decelerations of the impactor, did not cause greater discomfort. If discomfort is a valid proxy for increased injury risk in this experiment then in the tensed condition contusion injuries are less likely when tensed even with the higher peak impact forces.

Crisco et al. (1996) also found that in the tensed condition, for relatively more intense impacts on frogs, less energy was absorbed and less physical damage was done, but they also found that force was decreased. Their explanation was that the tensed muscle stopped the impact being dominated by the bone of the limb, with the stiffer and thickened muscle reducing the compression of the muscle tissue between the impactor and the bone. In this study the impacts were not severe enough that the bone would become a dominant factor and hence the likely reason for this difference in results between studies with regard to force.

Table 1
Impact force and energy absorption by the thigh for group means and standard deviations per condition, per drop height.

Drop Height (cm)	Tensed Peak Force (N)	Relaxed Peak Force (N)	Tensed Energy Absorbed (J)	Relaxed Energy Absorbed (J)
100	1411 ± 217	1187 ± 167	19.0 ± 5.3	24.3 ± 5.8
110	1224 ± 196	1204 ± 241	25.8 ± 4.9	28.0 ± 4.3
120	1280 ± 146	1311 ± 128	28.6 ± 4.6	30.3 ± 4.3
130	1367 ± 261	1137 ± 291	24.5 ± 6.6	36.2 ± 6.2
140	1383 ± 241	1283 ± 135	31.6 ± 7.4	34.1 ± 3.0
150	1507 ± 265	1215 ± 285	32.2 ± 8.5	40.1 ± 8.3
160	1478 ± 178	1385 ± 263	36.8 ± 3.9	39.4 ± 3.6

Table 2
Group mean impact characteristics for each drop height split into muscle condition. TTPF, time to peak force; CT, contact time.

Drop Height (cm)	Tensed			Relaxed		
	TTPF (ms)	CT (ms)	Comfort	TTPF (ms)	CT (ms)	Comfort
100	13.4 ± 1.5	43.4 ± 6.1	1.0 ± 0.07	16.0 ± 2.1	43.8 ± 5.8	3.9 ± 0.13
110	15.1 ± 2.3	45.7 ± 5.8	1.0 ± 0.08	16.5 ± 1.7	45.5 ± 7.0	3.8 ± 0.07
120	14.4 ± 2.3	42.2 ± 4.8	1.0 ± 0.03	14.3 ± 2.7	46.0 ± 6.6	3.8 ± 0.33
130	14.4 ± 2.3	45.6 ± 4.8	1.3 ± 0.14	16.0 ± 2.4	47.8 ± 8.2	4.1 ± 0.29
140	13.8 ± 2.4	43.6 ± 4.4	1.4 ± 0.14	15.3 ± 1.0	39.3 ± 6.7	4.1 ± 0.42
150	13.0 ± 3.9	41.7 ± 4.1	1.5 ± 0.15	15.2 ± 1.7	44.0 ± 9.9	4.8 ± 0.41
160	13.4 ± 2.2	43.1 ± 5.3	1.7 ± 0.14	14.9 ± 1.9	45.7 ± 8.9	4.9 ± 0.09
Mean	14.0 ± 2.4	43.7 ± 4.9	1.4 ± 0.28	15.6 ± 2.0	44.6 ± 7.6	4.2 ± 0.46

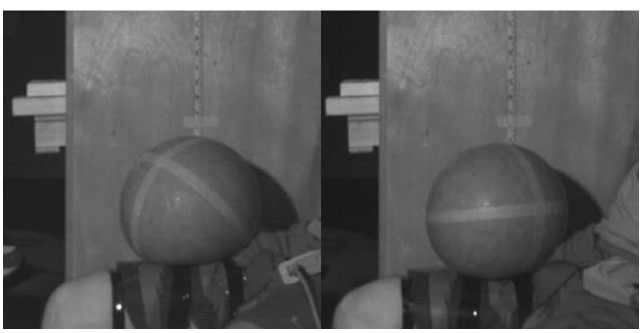

Fig. 3. Representative trials from a subject. Left is maximum compression during a tensed impact whilst right is maximum compression during a relaxed impact.

Elastic impact theory postulates that higher forces will cause higher internal stresses and subsequently more serious injury, and thus reducing forces during impacts is often a primary goal in injury mitigation practices. However, human impacts are complex as not only are limbs compound structures made of different materials but when muscle changes its structure to contract, it also alters its stress-strain properties. Activated muscle has a much higher stiffness along the line of the fibres than when relaxed, whereas transverse stiffness presents a more complex case and has been seen to decrease, not change much, or increase during electrically stimulated rigor (Sugi and Tsuchiya, 1982; Hatta et al., 1988; Tsuchiya et al., 1993).

If muscle injury is primarily caused by the crushing of soft tissue between the impactor and the underlying bone (de Souza and Gottfried, 2013) then there are a number of ways in which tensing muscle can ameliorate this. Increasing the thickness of the muscle under the impact, and increasing the stiffness of the muscle were described as possible mechanisms by Crisco et al. (1996). However, the actual thickness change can be small and impacts are mainly along the transverse axis where stiffness may not have changed, or may even have decreased, and these factors may not be sufficient to account for the large reductions seen in discomfort. Muscle tissue does not grow out from, or sit in a fixed position relative to the underlying bone but is effectively suspended like a thick net of fibres between tendons above the bone. This consideration of the anatomy and morphology of musculotendon units allows further mechanical methods for reducing compression during impact to be postulated. When relaxed, the net of fibres is slack and resting against the bone and so can easily be crushed up against it with only its inherent transverse stiffness to resist the compression. When tensed the net of fibres would not only have to be compressed, which may be no more difficult than when relaxed if dependent on the transverse stiffness, but depressed down as a whole onto the bone while it is under tension. This means deflecting the fibres and so the resistance to this would be dependent on the tension in the longitudinal direction, plus the geometry of the deflection, as seen in many effective cable tension support systems. An illuminating example, which can be readily felt in the arm of the reader, is shown in Fig. 4 which depicts the

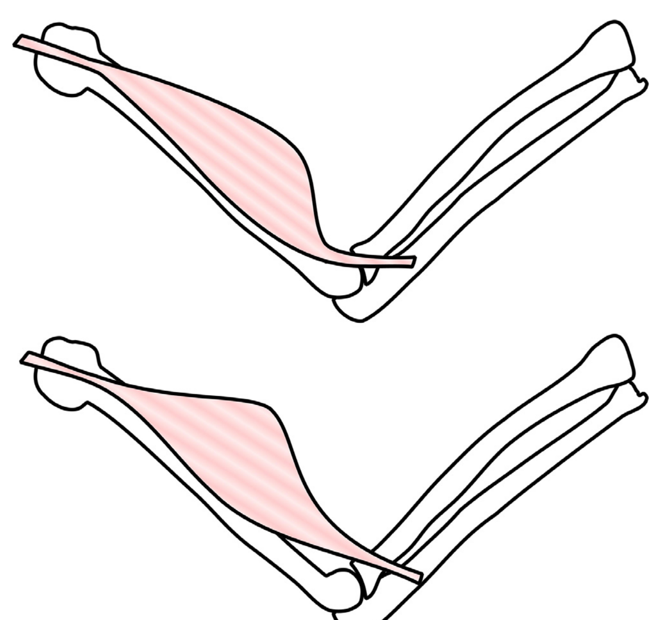

Fig. 4. A schematic of the biceps brachii for the same joint angle in relaxed, top, and tensed, bottom, conditions.

biceps musculotendon unit in relaxed and contracted conditions. In the relaxed condition the slack tendon, and tissue below it, can be easily pressed down into the humerus and with a forceful application it can become painful quickly. When isometrically tensed, requiring co-contraction of the triceps, the tendon can be felt stiffening and rising up, which alleviates the pain due to tissue compressing against the humerus. Although this easy to experience example is with the tendon the tendon is in series with the muscle belly so has the same force and lines of action. Another factor with this tensioned cable type resistance is that more of the muscle mass than just that directly under the impactor can be involved in the impact and so provides more inertial resistance to compression. This would hold as long as the time is sufficient for propagation of the forces along the fibres. Given the times to peak force found here, >10 ms, and the speed of sound in muscle being approximately 1540 m/s there is time for metres of muscle fibres to become involved.

These mechanisms could also explain that while at each drop height energy absorption was greater in relaxed muscle and correlated positively with discomfort for both muscle conditions, in tensed impacts even when more energy was absorbed by the thigh compared to relaxed impacts discomfort was still much lower. Increasing the effective mass of the impacted body reduces the energy per unit volume that is absorbed, and longitudinally stretching the stiffer activated muscle means more work needs to be done to stretch the tissue a given distance. A further possible non-standard mechanical factor worth considering is that by having the muscle resistance acting along the line of the muscle fibres, elongation by reversible, and less damaging, cross bridge detachment could occur to absorb some of the impact energy.

One limitation is that in this study, a subjective metric, perceived impact intensity, was used to infer an objective injury outcome (e.g. contusion) on a limited range of subjects. For an individual, especially those with long term experience of impacts, under the same impact conditions the more painful the impact the more likely it is to cause injury. However, injurious impacts can go unnoticed dependent on various psychophysical conditions at the time, such as inattention and the Gated Theory of pain perception (Melzack and Wall, 1965). In this study advanced knowledge and equivalent attention was present for all impacts so it is unlikely this would have altered pain perception between conditions. However, Gated Theory proposes that the central nervous system would be busy processing the electrical signals associated with muscle contraction, a larger fibre input, to properly assess the pain stimuli, a small fibre input. As such, the blocked stimuli may have led to lower perceived impact intensities during tensed muscle trials if this was in effect. It should also be noted that the F-scan pressure insoles only measure normal forces and so there is the limitation that not all impact force will have been determined. There could be slight differences in the impact force vectors between the tensed and relaxed conditions with the thigh muscle deforming differently. However, this difference was considered to be minimal, and along with the results not being depended on absolute force values should not alter the main outcomes of this study. A better dynamic calibration in the future may be to use a rigid ellipsoid fixed to the force plate with the F-scan sensors attached to this. Although repeated digitizing and averaging of points was used to get ball velocities and every effort was made to avoid including oblique impacts the use of a 3D system with higher resolution would be advisable in future studies.

Whatever its precise mechanism, it is clear that whether muscle is tensed or not has a significant and meaningful effect on injury risk during human on human type impacts. Unless the activation state of the muscle is known, using force is not a good indicator of pain, and thus likely not a good indicator of injury risk, and that energy absorption may be a more appropriate parameter to consider for these intensities of impacts onto human musculature. Given the large difference in perceived discomfort and the mechanical differences in a tensed condition from a relaxed condition, muscle tension state needs consideration for the development of more biofidelic impact models, both physical models and computer simulations.

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Disclosure of conflict of interest

There are no conflicts of interest.

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