# Effects of the abdominal belt on the reduction of spinal forces and muscle activities during extreme transits of high-speed craft

Zhenkai Zhao[1#], Leiming Gao[1], Benjamin Simpson[1], James Campbell[2], Neil J. Mansfield[1#]

1. Department of Engineering, Nottingham Trent University, Clifton Lane, Nottingham NG11 8NS, UK.

2. BMT, 1 Park Road, Teddington, London, TW11 0AP

#Corresponidng author: Neil J. Mansfield, neil.mansfield@ntu.ac.uk; Zhenkai Zhao, zhenkeizhao@gmail.com

## Abstract

Repeated high-g shocks and whole-body vibration (WBV) as experienced by operators of High-Speed Craft (HSC) increase the risk of fatigue, back pain, and acute and chronic injuries, especially in the lumbar region of the spine. Studies on abdominal belts have suggested their beneficial effects on lumbar torso stabilisation and back pain mitigation in both weight lifting and HSC scenarios. This paper presents a human musculoskeletal model to simulate belt effects for occupants on HSC under high-g shocks. Parameters included the shock severity with peak acceleration ranging from 3g to 10g, human dimensions and muscle strengths, the belt width and belt forces are investigated. The results show an average of 120% increase in the intra-abdominal pressure (IAP), a 9% reduction in the transverse abdominis activities, and a 12% reduction in the spinal compressive force at the L4/L5 joints when the abdominal belt is added to the human model. In conclusion, wearing an abdominal belt significantly assists abdominal muscles and maintains a solid core during intense WBV generated in different shock severity levels. It may cause a small negative influence on the neck region with a 2.4% increase in the shear force at the C4/C5 joints.

**Keywords:** Abdominal belt, whole-body vibration, intra-abdominal pressure, spinal unloading, muscle activity.

## 1. Introduction

Passengers and crew of small fast boats (High-Speed Craft, HSC) can experience severe repeated shocks and whole-body vibration (WBV) (Halswell, Wilson et al. (2016). The harsh sea environment for HSC can lead to detrimental effects on human beings, including motion-induced fatigue and the risk of acute and chronic injury (J.Mansfield 2005, Dobbins, Rowley et al. 2008, Myers, Dobbins et al. 2011). Low back musculoskeletal injury and pain are the most common symptoms, with apparent muscle, ligament, and bone pains in the lumbar region (Bridger 1999, Bartleson 2001, Bridger and Padden 2003). Existing approaches to mitigate the harmful effects of marine shock and vibration (S&V) include limiting the craft speed, improving the hull and seat design to absorb the forces and reducing acceleration levels (Townsend, Coe et al. 2012). (Garme, Burstrom et al. 2011, Myers, Dobbins Td Fau - King et al. 2012).

Although the absorption of S&V with advanced seat and hull structure is effective, the increased weight, complexity and cost cannot be neglected, and residual risks remain. Therefore, the development of an abdominal/back belt was considered. Initially, similar belts were designed for heavy weight lifters and athletes. Belt effects on the intra-abdominal pressure (IAP) were measured and compared to the case without the belt. (Grillner S Fau - Nilsson, Nilsson J Fau - Thorstensson et al. , Morris, Lucas et al. 1961). Based on the measured data, there was a significant increase in IAP with the belt to stabilise the lumbar region. Similar conclusions were made by Miyamoto *et al.* (Miyamoto, Iinuma et al. 1996); an increase in intradiscal pressure was found after the belt was applied, proving that wearing abdominal belts may contribute to the stabilisation during lifting exertions. Considering the belt effects on spinal loading reduction, Jeffrey *et al* (Woldstad and R. Sherman 1998) measured the L4/L5 spinal compressive force. They found that the abdominal support belt can reduce the spinal compressive force. However, controversial results were gained by Ivancic *et al*, who found that wearing an abdominal belt did not mitigate the joint compression forces at L4/L5 (Ivancic, Cholewicki et al. 2002).Their biomechanical model didn't consider the role of transverse abdominis on spinal unloading, which might lead to the controversy. The study of belt effects was not limited to weight lifting or daily physical activities, but also include experiments under severe conditions. The Royal Navy carried out a trial of back belts for crew members on rigid inflatable boats at different levels of sea wave conditions, participants were rated with back pain magnitude ranging from 1 to 10g and found that the wearing of belts can reduce back pain in the boat crews; few unwanted side effects were observed as well (Bridger and Padden 2003).

The abovementioned methods mainly consist of direct experiments on the human being; IAP and muscle forces were measured with transducers on human skin and electromyography. However, high costs, potential injuries to the participants, and long duration for some specific experiments are deficiencies for human experimental tests. It is also difficult to perform experiments with subjects exposed to extreme shock such as those experienced on HSC. A musculoskeletal model can provide an inexpensive and efficient method to determine features such as muscle activation patterns, joint forces and torques, contributions of passive and active stiffness elements, and optimal posture. It can be used to compare the motion of human beings with and without lower back pain and evaluate belts' effects. Musculoskeletal models were developed with existing anthropometric data to estimate the muscle functions such as muscle force production and performance (de Zee, Hansen et al. 2007, Christophy, Faruk Senan et al. 2012)However, these models either excluded or simplified IAP functions on the spinal unloading and stability.

To investigate the IAP effects in the musculoskeletal model, Arshad *et al.* (Arshad, Zander et al. 2016) built a standing human body model. They found that modelling with IAP leads to closer results to the experiments. Similar research was carried out by Liu *et al.* (Liu, Khalaf et al. 2018), but a coupling of musculoskeletal and Finite-element (FE) models was applied. Using the coupled model, the spinal components' load-sharing effects were studied. Besides, Liu *et al*. (Liu, Khalaf et al. 2019) assessed the IAP effects with the optimisation of neglecting the IAP term but keeping the activation of abdominal muscle. As a result, the inclusion of IAP reduced global muscle forces, the disc loads and intradiscal pressure. These models have reinforced the role of IAP and provided essential information for accurately modelling the lumbar spine.

Inspired by the previous research, this paper presents a study of the effect of a virtual belt on a seated human model experiencing severe S&V. Results included IAP, muscle activity, and spinal compressive are obtained under different sea states, anthropometric dimensions, and belt forces. The results of cases before and after the belt is applied are compared. Relationships between muscle strength and belt effect are discussed as well.

## 2. Methods

### 2.1. Musculoskeletal modeling

The seated musculoskeletal model was developed using AnyBody modelling system (AnyBody Technology Company, Denmark) a commercial software for musculoskeletal simulation analysis based on the inverse dynamics method, as shown in **Fig.1A**. The model was originally developed by de Zee *et al* (de Zee, Hansen et al. 2007) consisting 7 rigid bodies and 8 types of muscles of the back and abdomen. It has 18 degrees of freedom, allowing extension-bending movement in the sagittal plane and lateral bending in the frontal plane, and rotation in the transverse plane, which is adequate to simulate the spinal compressive force with justified references in the abovementioned paper. The seated human model in AnyBody consisted of skull, spine, arms, legs, pelvis, and feet. The spine was divided into cervical, thoracic and lumbar regions, as shown in **Fig.1B**. The cervical region included eight vertebra segments ranging from C0 to C7. The thoracic spine was modelled as one rigid segment with vertebrae noted as T1 to T12 constrained with spherical joints. The lumbar spine had five rigid vertebrae noted as L1 to L5, as well as the sacrum S1 This structure was rigidly connected to the pelvis. The cervical and lumbar regions were connected by joints at T1C7 and T12L1. The intervertebral spherical joints allow three degrees of freedom and can provide reaction forces. In the current model, ligaments were not included, which might cause an underestimate of the calculated spinal force. However, previous study has assumed that no significant contribution to torque production from ligament, (Daggfeldt and Thorstensson 2003). The assumption was made given that if some ligaments can contribute like the iliolumbar ligament much more precise methods are required to determine the skeletal geometry to estimate the forces. Since more information such as slack lengths of the ligament is not available, the current model could give inaccurate results to specific human bodies. Therefore, the model will not take ligaments into account. The spinalis and transverse abdominis groups considered in the model are shown in **Fig 1C**.

To solve the redundancy problem that existed in muscle recruitment, AnyBody uses static optimisation with the assumption that muscle activity follows the optimisation principle (Damsgaard, Rasmussen et al. 2006), and expressed with the following form:

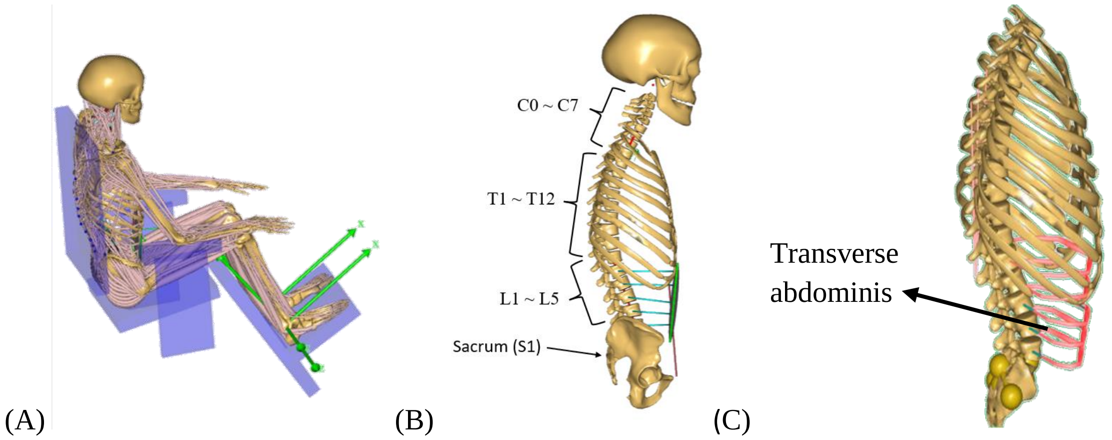



Figure 1. (A) Seated human model used in *AnyBody*; (B) The cervical, thoracic and lumbar regions of the spine with noted vertebrae segments; (C) Transverse abdominis specified in red.

$$Minimize\ \ G(f^{M}) \tag{1}$$

$$Cf = F; f = [f^{(M)T} f^{(R)T}]^{T} \tag{2}$$

$$0 \leq f_i^{(M)} \ll N_i, i \in \{1, \dots n^{(M)}\} \tag{3}$$

In Equation (1), $G$ is the objective function, assuming that muscle force is optimised with Central Nervous System (CNS) criterion. $G$ is minimised concerning all unknown forces (*f*) in the problem, mainly including muscle force $f^{(M)}$, and joint reaction force $f^{(R)}$. Equation (2) is the constraint function in the optimisation problem derived from the dynamic equilibrium equation. $C$ is the coefficient matrix, and $F$ contains all known applied loads and inertia forces. The superscript $T$ denotes the transformation matrix. Equation (3) defines the bounds of any muscle force $f_i^{(M)}$ between zero and the maximum strength of the $ith$ muscle $N_i$. The physical characteristics of the muscle forces are limited by non-negativity constraints, which means the muscle can only pull, instead of push.

In addition, AnyBody uses a minimum-maximum fatigue algorithm to address the muscle recruitment issue (Rasmussen, Damsgaard et al. 2001). This method assumes that all the muscles respond to the external load in a coordinated manner (Wang, Huo et al. 2022). The maximum muscle activity is minimised and equals the minimum muscle force, as expressed in Equation (4).

$$\min G\left(f^{(M)}\right) = \max\left(\frac{f_i^{(M)}}{N_i}\right),\ \ i \in \{1, \dots n^{(M)}\} \tag{4}$$

### 2.2. Shock pulse generation and seat boundary constraints

To mimic the realistic S&V caused by the hull's impact on the sea surface, vertical vibrations represented by different magnitudes of vertical acceleration were produced with a digital model in MATLAB on the deck and the seat (Olausson and Garme 2014) on the deck (floor) and the seat (surface). As shown in **Fig. 2**, an example of the actual shock pulse on the seat was simulated as a half-sine pulse (red curve) with a magnitude of 75 $m/s^2$and period of 0.2 s; after fitted to sine function (black curve), describing the vertical acceleration exerted on the seat surface used in the simulation.

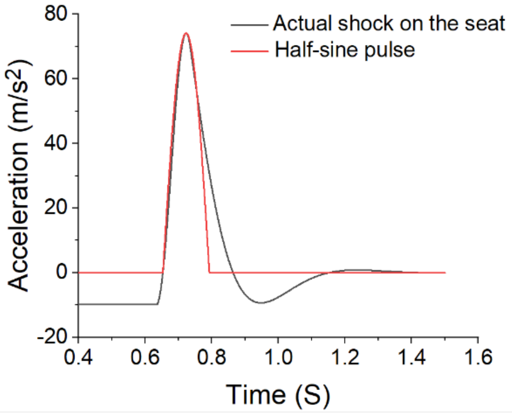


Figure 2. Half-sine pulse for simulation and the actual shock pulse on the seat.

Based on the original template *SeatedHuman* from the *AnyBody Manages Model Repository v. 2.3 (Skals, Blafoss et al. 2021)*, The seated model mainly consists of the backrest, seat pan, armrest, leg rest, headrest, and footrest as boundaries, as shown in **Fig. 1(A)**. There is a support muscle with 10000N strength as contact elements for all the boundaries. Details of boundary constraints at different locations in the human model are shown in **Table 1**. The frictional coefficient of 0.5 was setup in AnyBody for all the boundaries except for arm and armrest for which it was 0.2. The seat was not in an upright position, with the inclination angle of 10 degrees pointing upwards, based on the real seat used on HSC. The boundary constraints were used for all cases to ensure the same influences on the results.

Table 1. Boundary constraints of the seated human musculoskeletal model

| Locations | Boundary constraints |
| --- | --- |
| Back and backrest | Friction coefficient 0.5; thoracic T2, T6, T9, T10 and lumbar spine L1-L5; 10 support points in total |
| Thing and seat pan | Friction coefficient 0.5; two support nodes on front and back of thigh symmetrically; Four support nodes on left and right of ischia symmetrically; 12 support nodes in total |

| | |
|---|---|
| Foot and footrest | Friction coefficient 0.5; two support nodes on heel and toe symmetrically; 4 support nodes in total |
| Leg and leg rest | Friction coefficient 0.5; two support nodes on shank symmetrically |
| Arm and armrest | Friction coefficient 0.2; two support nodes on Ulna symmetrically |
| Seat pan and backrest | Revolute joint |
| Backrest and headrest | Revolute joint |
| Seat pan and leg rest | Revolute joint |
| Leg rest and footrest | Prismatic joint |

### 2.3. IAP and Abdominal belt

The model contained 5 rigid vertebrae, the pelvis, and the corresponding muscles in the lumbar region. As shown in **Fig. 3(A)**, five vertebrae segments numbering from L1 to L5 were distributed in the lumbar region. Based on the *SeatedHuman* model in AnyBody, connections among disk, vertebrae and buckle were spherical joints. The IAP model is an embedded model in the software and was included with one rigid buckle providing attachments for the different types of abdominal muscles, and five rigid artificial disks connected to the buckle, providing a structure for the transverse abdominis to be attached. The transverse abdominis is responsible for generating the IAP by controlling the movement of the artificial segment (shown in **Fig. 3(B)**) when the muscle is activated during contraction or postural changes.

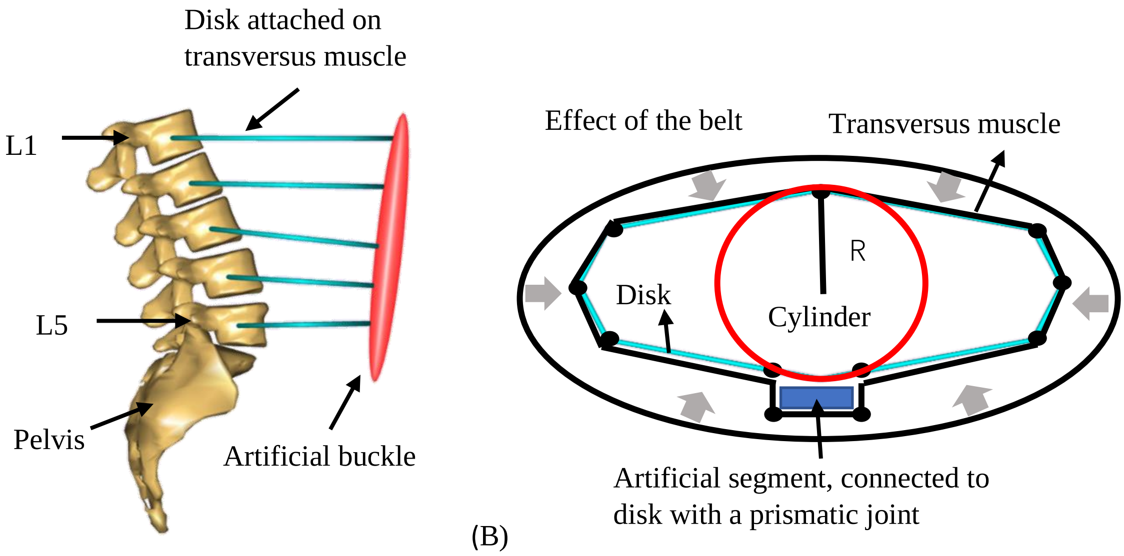
The volume of the abdominal cavity was estimated by a cylinder with a radius *R* and height H as shown in **Fig. 3(B)**, the movement of the artificial segment, together with changing distance between the thorax and pelvis, can change the volume V of the abdominal cavity. Using this estimation method, the volume can be linearised as a function of the radius and height as expressed in Equations (6)-(8):

$$V = \pi R^2 H \quad V = V_0 + \frac{dV}{dR}\Delta R + \frac{dV}{dH}\Delta H \tag{6}$$

$$\frac{dV}{dR} = 2\pi R H \tag{7}$$

$$\frac{dV}{dH} = \pi R^2 \tag{8}$$

IAP is generated during the change of the abdominal volume by an artificial muscle, with a maximum value of 26.6kpa (Essendrop 2004). The IAP artificial muscle is within the muscle recruitment system in AnyBody and applies force in the transverse plane to balance the transversus muscle forces to establish equilibrium. In this model, the stretchy belt performs as an artificial muscle that provides a squeezing force pointing inwards to the cylinder to stabilise the lumbar region and generate the IAP. The squeezing force behaves like the transverse abdominis to generate IAP during contraction or postural changes. Therefore, the belt is assumed as either an extra muscle in the transverse plane or it thickens the existing transverse abdominis to generate higher IAP which has been validated by previous research (Nowakowska-Lipiec, Michnik et al. 2020). The muscle force acted uniformly around the disk to represent the belt effect. The belt was applied to cover the five vertebrae (L1-L5) in most cases unless the parametric study section on the belt width.




(A) (B)

Figure 3. (A) Side view of the lumbar region structure with disks (cyan line) and buckle (red segment). (B) Top view of the abdominis cross-section.

**2.4. Parameters of the human body**

The human model was tested under various sea levels with peak acceleration ranging from 3g to 10g. Besides, subjects with 5$^{th}$, 50$^{th}$, and 95$^{th}$ percentile anthropometric dimensions on HSC were studied based on documents provided by the UK Ministry of Defence (2016). The AnyBody software has defined mass attributes for each body segment (bones and muscles) based on human anatomy, therefore, only the height and weight of the subjects need to be provided, and the software can allocate human mass to each body segment in proportion (Wang, Huo et al. 2022). The anthropometric dimensions of body weight and height are shown in **Table 2**. For a given belt force, the equivalent belt pressure was calculated by dividing the surface area of the abdominal cylinder by the total belt force ($F_{belt}$ ) as expressed in Equation (9), The different level of belt forces and corresponding equivalent pressure are categorised in **Table 3**. The belt force of 700N was selected with equivalent pressure of 2845 Pa, which is within the comfort zone of the human body (Lim and Davies 2014).

$$Pressure = F_{belt}/(2\pi RH) \tag{9}$$

Table 2. Body weight and height for different anthropometric dimensions of the male crew.

| | 5th percentile | 50th percentile | 95th percentile |
|---|---|---|---|
| Body weight (kg) | 69 | 81 | 96 |
| Body height (cm) | 169.3 | 178.1 | 189.8 |

Table 3. Belt forces and the equivalent uniform pressure exerted on the lumbar.

| | 1 | 2 | 3 | 4 | 5 | 6 |
|---|---|---|---|---|---|---|
| Belt force (N) | 100 | 500 | 700 | 1000 | 1500 | 2000 |
| Equivalent pressure (Pa) | 406 | 2032 | 2845 | 4065 | 6097 | 8130 |

2.5 Sea levels and the worst case

Subjects with different anthropometric dimensions experiences a wide range of sea wave levels (shown with **Table 4**). Passengers on the high-speed craft can experience vertical acceleration ranging from 3g to 10g according to the US DOD laboratory test requirements for marine shock isolation seats (Michael R. Riley, Kelly D. Haupt et al. 2018). Considering the actual environments, the worst case is defined with peak vertical acceleration, which is 10g in the paper for subjects of different anthropometric dimensions (studied in section 3.3).

Table 4. Severity threshold for different sea levels represented by peak vertical acceleration with constant impact duration.

| Threshold level * | Peak acceleration (g) |
|---|---|
| 6 | 10.00 |
| 5 | 8.00 |
| 4 | 6.00 |
| 3 | 5.00 |
| 2 | 4.00 |
| 1 | 3.00 |

**Threshold level defined by US DOD Laboratory Test Requirements for Marine Shock Isolation Seats***3.**

## Results

In this section, we firstly conducted a model validation according to two relevant previous studies, one from experimental tests and the other from numerical simulations. Then, several parametric studies on the magnitude of accelerations, the human body dimensions, belt forces and belt covering width and muscle strengths were presented in the following subsections. In the last subsection, we also discussed the effect of the belt on the cervical region, which is of concern as a critical part of the human body.

The direct results obtained from the human MSK model included the muscle strength and reaction forces at any joints. Four parameters were derived to evaluate the belt effect, IAP, transverse abdominis activity (TrA), the compressive forces at joint L4/L5 in the lumbar region and the shear forces at joint C4/C5 in the cervical region. IAP is closely related to the stabilisation of the lumbar region against external loading and its definition is described in Section 2.3. TrA is normalised by dividing the actual muscle strength by the maximum muscle strength of the transverse abdominis muscle group. Maximum muscle strength is an inherent property of the muscle, with a constant value of 246 N for transverse abdominis. Muscle activity can represent muscle recruitment; therefore, the reduction of TrA indicates the potential mitigation of muscle fatigue. The lumbar compressive force is a reaction force between adjacent vertebrae L4 and L5; reducing this force is a benefit to reduce the impact on the spine. Severe sea states have been known to cause damage to the neck region of seated HSC passengers. The cervical shear force at C4 and C5 is a parameter to evaluate the risk that could increase damage in the neck region.

### 3.1. Validation for the seated human musculoskeletal model

Considering the lack of research on IAP measurements and spinal compressive force under high acceleration amplitude, we carried out the validation with a seated car driver under small magnitudes of vertical acceleration. Parameters of the human body, seat and shock were chosen based on two literature papers (Verver, van Hoof et al. 2003) and (Li, Zhang et al. 2015) as listed in **Table 5**. We firstly compared the spinal compressive forces at lumbar disc sections between L1/T12 and S1/L5 with results from Verver's study. As shown in **Fig. 4**, the agreement was achieved in the predicted forces with a maximum of 18% differences and the same trend of the spinal compressive force decreasing from L1/T12 to S1/L5 was found as in the literature. The differences in the value of forces were mainly because of the means to obtain the compressive forces: experimental measurements in the literature versus numerical simulations with MSK models.

Table 5. Parameters of the human body, seat and shock in a seated driver model.

| Case No. | Body Height (cm) | Body Weight (kg) | Backrest inclination angle (degree) | Seat inclination angle (degree) | Frequency (Hz) | Vertical Acceleration (m/s^2) | Vertical motion amplitude (mm) |
|---|---|---|---|---|---|---|---|
| 1 (Verver et al.) | 175 | 75 | 7.3 | 9.8 | 8 | 0.4G | N/A |
| 2 (Li et al.) | 170 | 80 | 5-30 | 10 | 1-10 | N/A | 1.5 |

In the second validation case, we compared the maximum muscle activity in the lumbar region with a range of vertical vibration frequencies and backrest inclination angles according to Li's results. As shown in **Fig. 5**, the predicted muscle activity showed great consistency with less than 5% relative errors, compared to the literature.

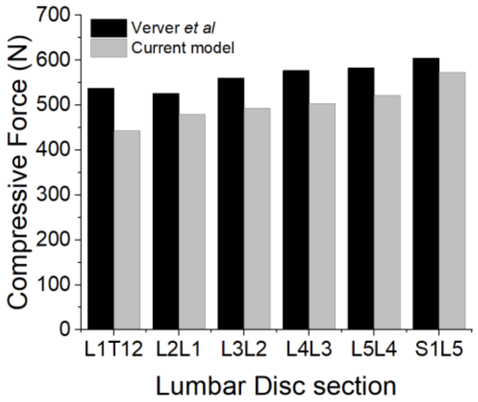
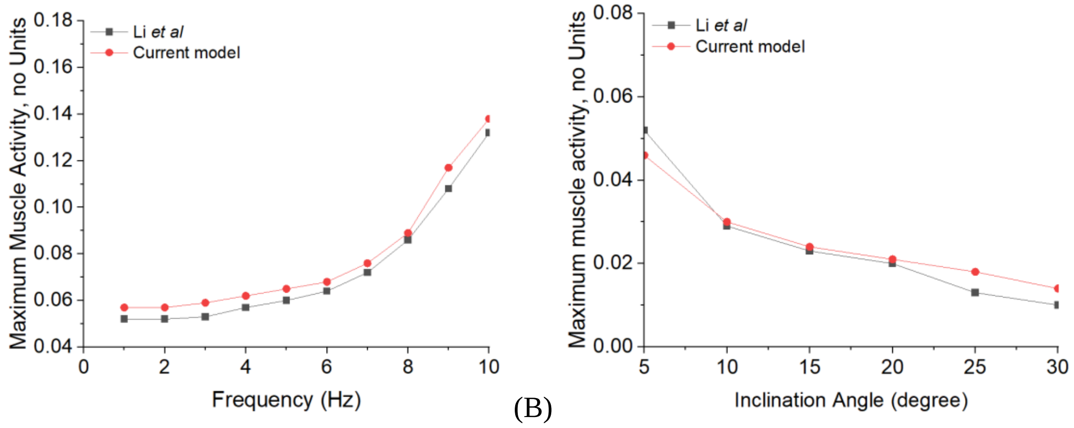



Figure 4. Validation results for the MSK model case 1: compressive forces for different lumbar disc sections.




(A) (B)

Figure 5. Validation results for the MSK model case 2: the maximum lumbar muscle activity against (A) vibration frequencies; (B) backrest inclination angles.

### 3.2. Parametric study on vertical accelerations

The magnitude of acceleration indicates the severity level of sea states. The peak vertical acceleration ranging from 3g to 10g was studied. The 50% percentile male model was selected for all cases in this section. For the most severe level of 10g vertical acceleration, the results of the three parameters to evaluate belt effects during the shock period are compared in **Fig. 6.** We found a significant increase in IAP (approximately 120%) and a decrease of TrA (approximately 18%) after the belt was applied. In addition, an approximate 12% reduction of spinal loading at the L4/L5 joint was found, supporting the assumption that the abdominal belt benefits passengers with a relief of the spinal load.

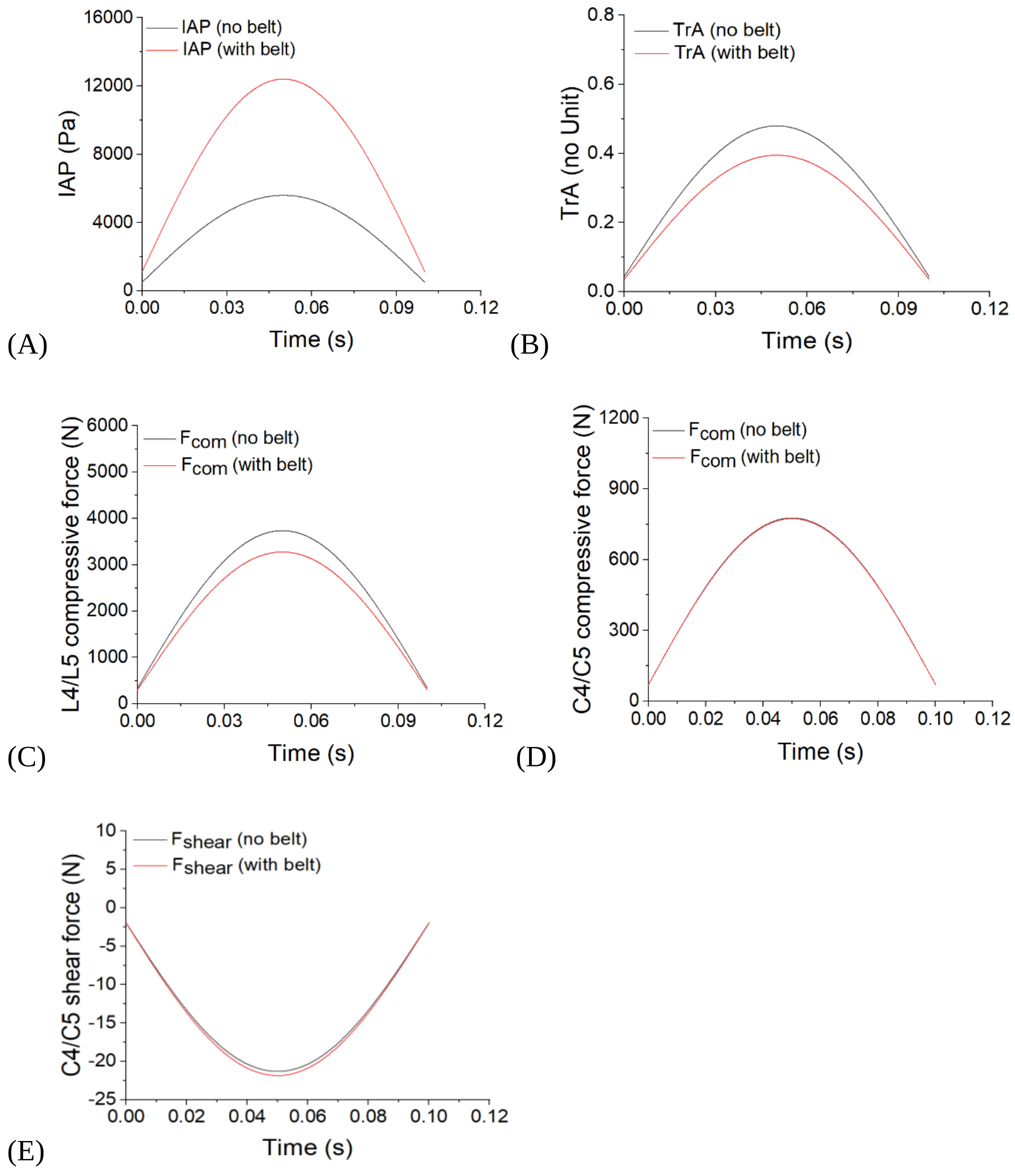



Figure 6. Belt effects evaluation with three parameters (A) intra-abdominal pressure; (B) transverse abdominis activity; (C) L4/L5 compressive force; (D) C4/C5 compressive force; (E) C4/C5 shear force during the shock time under 10g vertical acceleration.

The variations of the maximum IAP, TrA and L4/L5 compressive force against the acceleration are shown in **Fig. 7**. As expected, the belt was beneficial within the entire range (3g-10g) of accelerations, suggesting that the squeezing force provided by the belt can effectively support passengers on HSC in various sea states.

The results of C4/C5 joint forces (**Fig. 6D-E** and **Fig. 7D**) implied that the belt does not affect the compressive force in the cervical vertebrae but has a marginally negative effect on cervical vertebrae shear force (2.4% increase when wearing the belt). However, on balance, this is a much smaller impact than the positive effect the belt has on spinal unloading (12.3%).

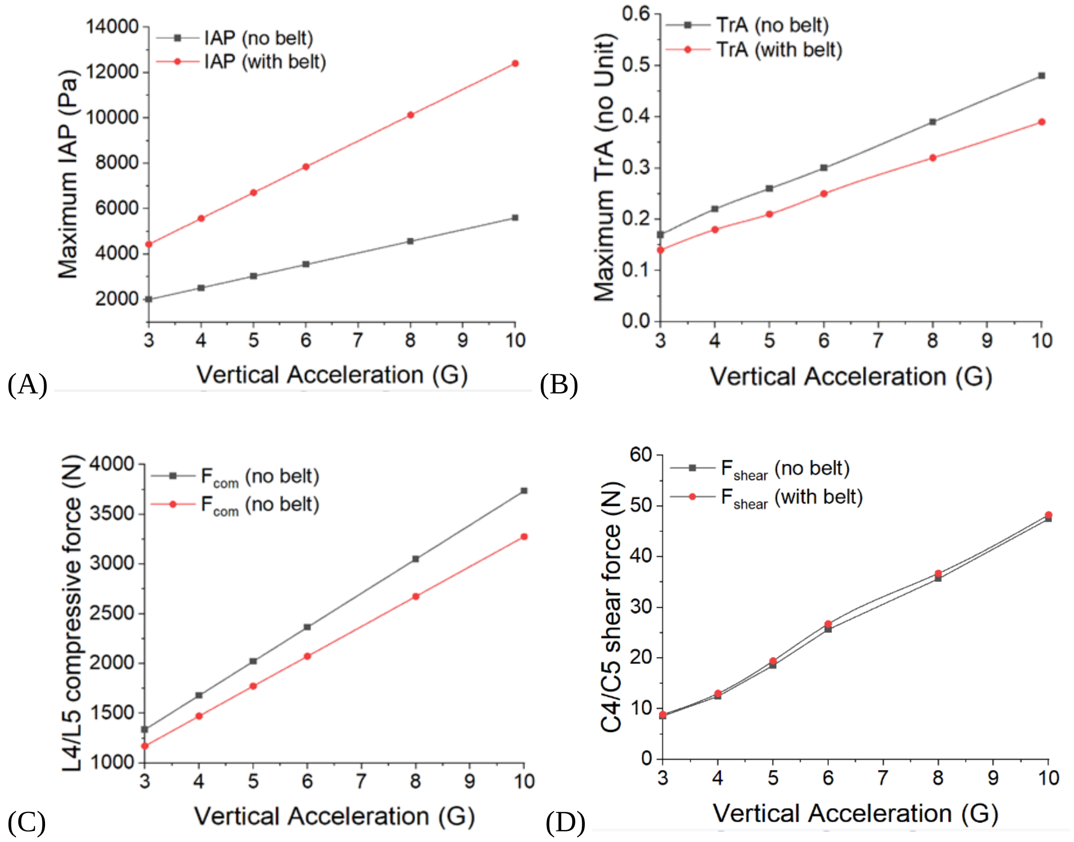



Figure 7. Belt effects under different vertical accelerations. (A) maximum intra-abdominal pressure; (B) maximum transverse abdominis activity; (C) maximum L4/L5 compressive force; (D) maximum C4/C5 shear force.

### 3.3. Parametric study on anthropometric dimensions

The anthropometric dimensions mainly included the human body weight and height, as described in Section 2.4 Table 2. Results of the three human dimensions, 5$^{th}$, 50$^{th}$ and 95$^{th,}$ under 10g vertical acceleration are presented in **Fig. 8**.

The belt provided an overall positive effect for different anthropometric dimensions, with approximately 107%-134% increase in IAP, 7%-10% reduction in TrA and 10%-12% reduction in L4/L5 forces. But the effects tended to be slightly smaller for people with 95$^{th}$ percentile dimensions compared to the other two groups (approximately 10% versus 12% reduction in L4/L5 compressive forces).

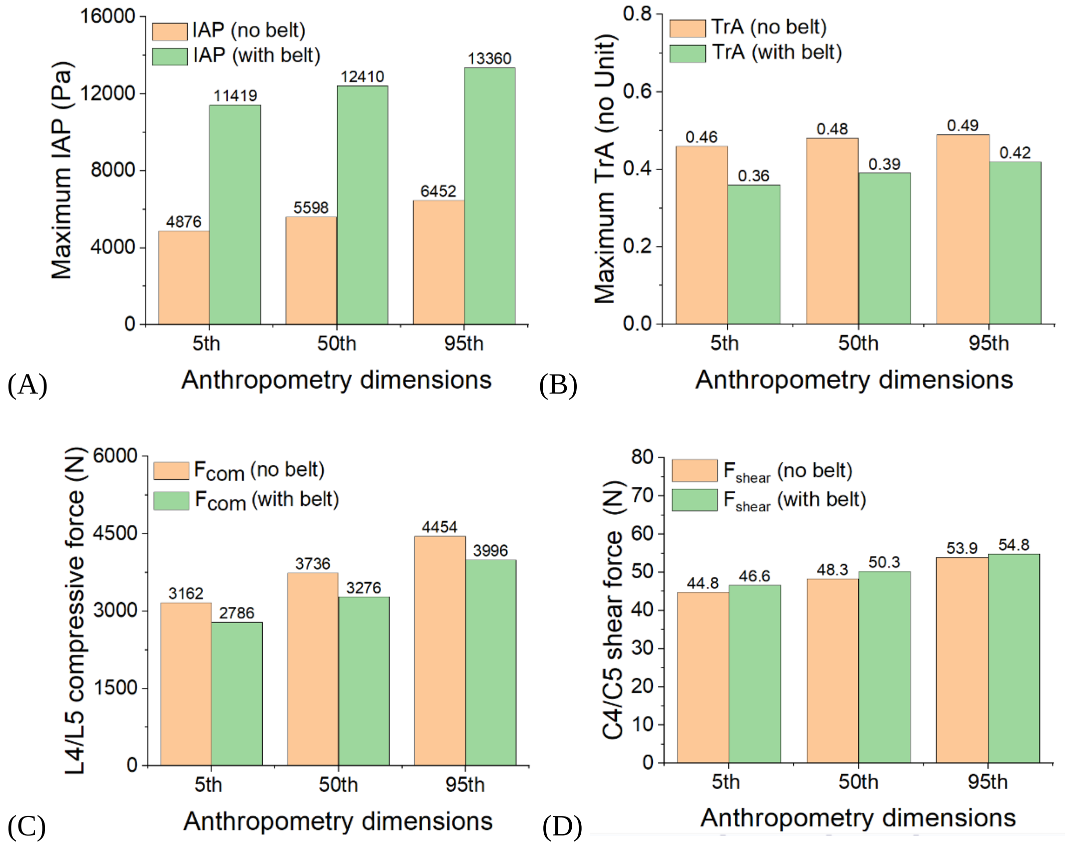



Figure 8. Belt effects for different anthropometric dimensions. (A) maximum intro-abdominal pressure; (B) maximum transverse abdominis activity; (C) maximum L4/L5 compressive force; (D) maximum C4/C5 shear force.

### 3.4. Effects of the belt force

This section studied the belt forces ranging from 100 N to 2000 N, as described in Table 3, and the belt width covering from one segment to full five segments between L1 and L5 in the lumbar region. It is obvious that larger belt force led to better performance, i.e., IAP increase, TrA decrease and joint compressive force decrease. However, there was a tendency that as the force increased, the effects of the belt approached a plateau (**Fig. 9A-C)**. This finding would be much clearer if the changes in IAP and L4/L5 joint force were plotted as per Newton, as shown in **Fig. 9E and F**, that both the IAP increment and compressive force reduction reached their maximum values when the total belt force was between 500N to 1000N.

Regarding the cervical region, there was a slight increase of shear force at the C4/C5 joint in the cervical region after the belt was applied (**Fig. 9D**). However, once the belt force exceeded 1000N, the shear force had a constant 6%-7% increase, which means the appropriate belt force needs to be carefully selected to avoid high risk in the cervical joint damage.

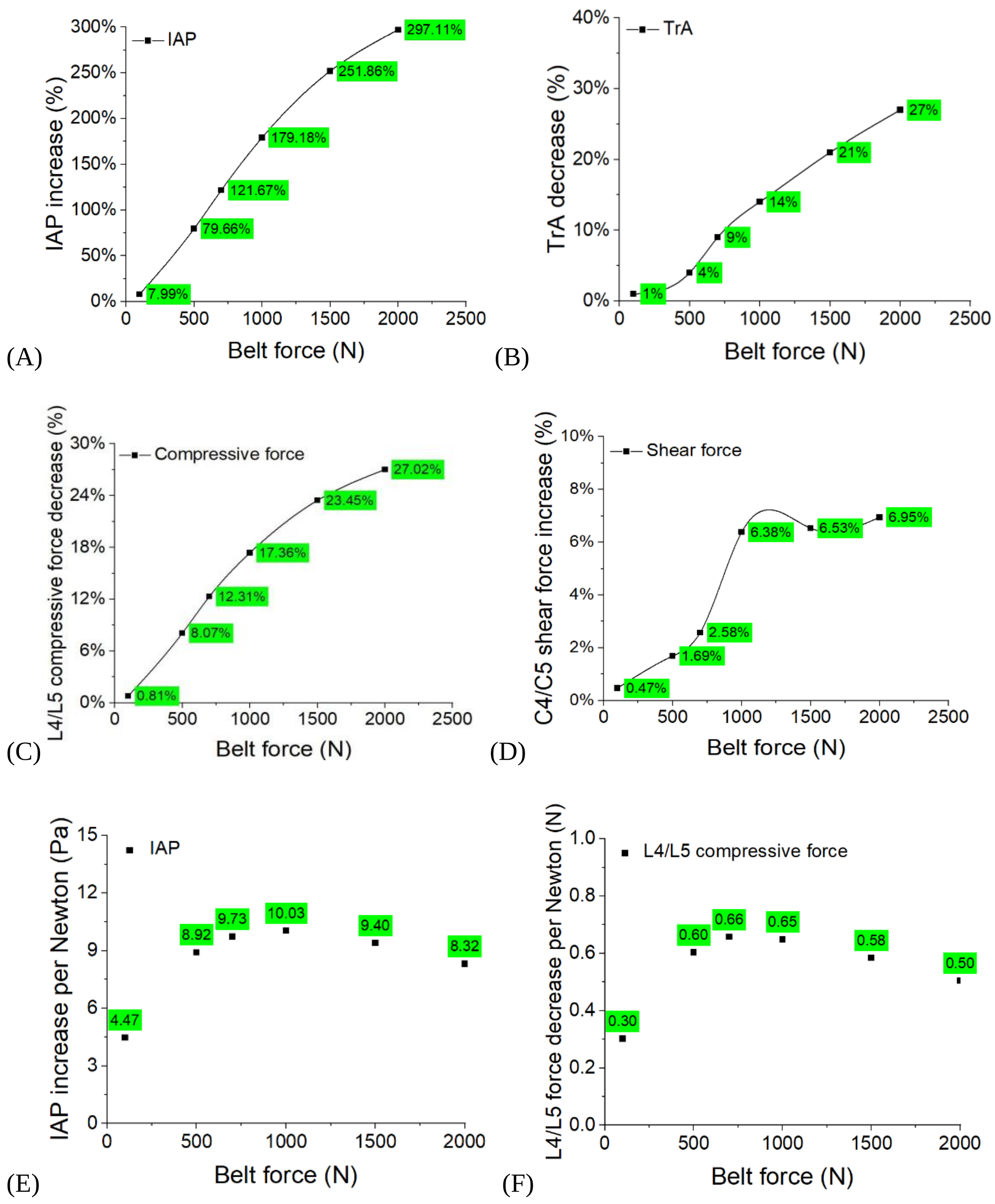
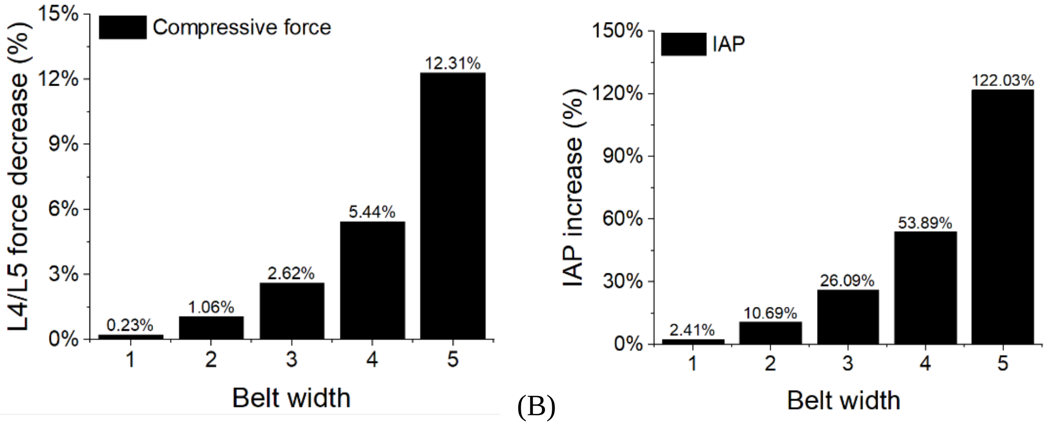



Figure 9. Effect of the belt force on(A) Intra-abdominal pressure; (B) Transverse abdominis activity; (C) L4/L5 compressive force; (D) C4/C5 shear force; (E) Normalised IAP; (F) Normalised L4/L5 compressive force.

### 3.4. Effects of the belt width

In previous sections of results, the belt covered all five vertebrae segments in the lumbar region. In this section, narrower belts that covered less number of vertebrae segments were investigated to evaluate the influences of the belt width on belt performance. The belt width levels were represented by the number of segments covered by the abdominal belt, i.e., from 1 to 5. The results showed that a level 5 belt width, i.e., a belt that covers the entire lumbar region, had the best belt performance in terms of reducing the spinal joint forces and increasing the IAP as shown in **Fig. 10.** The human dimension of 50$^{th}$ percentile and peak acceleration of 10g were adopted in this sub-section.




(A) (B)

Figure 10. Effect of the belt width on (A) intro-abdominal pressure; (B) L4/L5 compressive force.

### 3.5. Effect of the muscle strength

The transverse abdominis and the spinalis muscles are the relevant muscle groups that contribute to the IAP and spinal loading. Therefore, the diversity of their strengths was investigated in this section. Six simulation cases with the muscle groups weakened to 70% and strengthened to 120% of the original strengths are presented in **Table** 6. The weakened and strengthened muscle groups represent people with weakened and stronger muscles, respectively. The muscle was weakened or strengthened by changing the cross-sectional area of muscle fibre in the AnyBody human model.

The influence of muscle strengths was first studied without the belt, as shown in **Fig. 11**. It was found that larger transverse abdominis led to higher IAP. A weakened muscle was compensated by higher activity in other muscles to reach a balance for the activity of the entire muscle groups. Also, the compressive force increases once the muscle has been weakened. This is reversed for strengthened muscle groups. **Table 7** shows the belt effects in the six cases of different muscle strengths. The key finding was that in no muscle configuration did the belt make the lumbar loading worse. And there was only a difference in the level of the benefit. For example, the belt tended to perform better for people with weakened transversus muscles than those with stronger ones, with the reduction in L4/L5 compressive forces to 13.2% versus 11.4%. This conclusion is in line with the parametric study on anthropometric dimensions in Section 3.3.

Table 6. The muscle strength modification for the transversus and spinalis muscle groups

| Case No. | Muscle groups and the modification | % of original value |
|---|---|---|
| 0 | Original setting (Original) | 100% |
| 1 | Transversus muscle weakened (TW) | 70% |
| 2 | Spinalis muscle weakened (SW) | 70% |
| 3 | Both muscle groups weakened (SW&TW) | 70% |
| 4 | Transversus muscle strengthened (TS) | 120% |
| 5 | Spinalis muscle strengthened (SS) | 120% |
| 6 | Both muscle groups strengthened (TS&SS) | 120% |

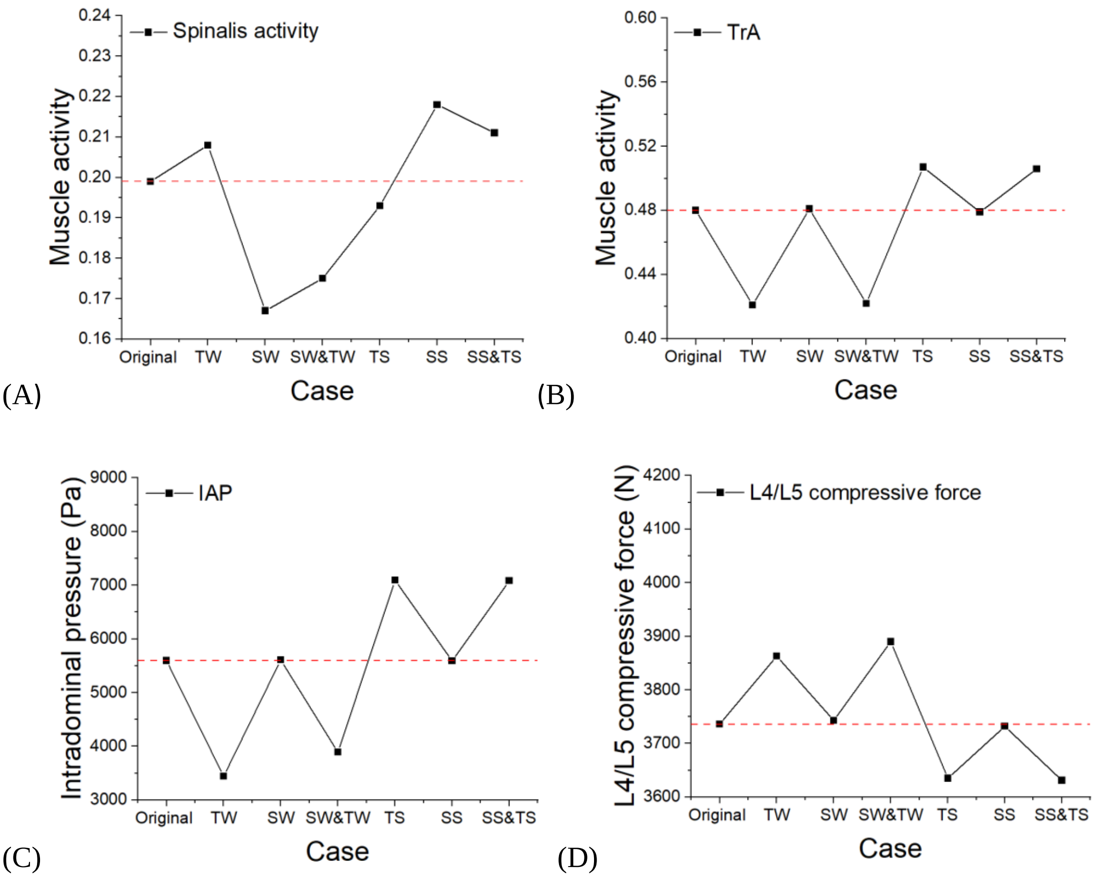



Figure 11. Prediction of IAP, TrA and spinal forces in six cases of various muscle strengths of the transversus and spinalis muscle groups.

Table 7. Influence of the belt in the cases of various muscle strengths

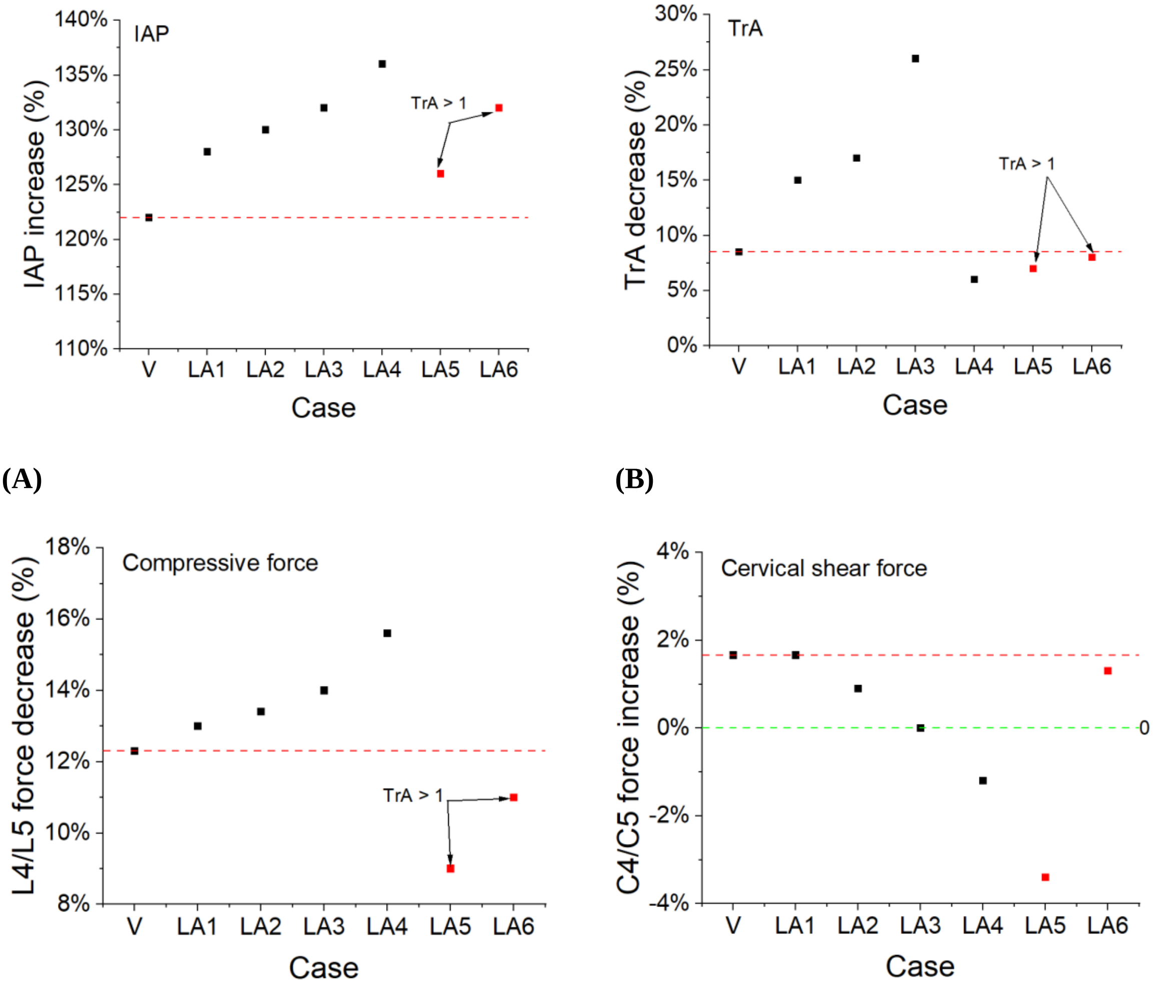
| | | | | | | | |
|---|---|---|---|---|---|---|---|
| TrA decrease | 8.50% | 7.50% | 8.50% | 7.50% | 9.00% | 8.50% | 9.00% |
| IAP increase | 121% | 220% | 121% | 220% | 88.3% | 121% | 88.3% |
| L4/L5 force decrease | 12.3% | 13.2% | 12.3% | 13.2% | 11.4% | 12.3% | 11.4% |

### 3.6. Effect of the belt under multi-dimensional vibration

Accelerations composed of both vertical and lateral directions in the frontal plane were studied to simulate the impact of side landing scenarios of the HSC.  As shown in **Table 7**,The peak magnitude of the resultant acceleration remained at 10g in all conditions. . The effects of the belt were depicted in **Fig. 12**. In general, for the same belt force, the belt shows a better protection effect in higher lateral acceleration. However, this is not applicable for extremely large lateral acceleration (such as case LT5 and LT6), which can lead to the muscle activity larger than 1. beyond the threshold, the muscle is not capable of responding to the shock, indicating a limiting value for shock exposure laterally. As a consequence, the effects of the belt were significantly reduced, represented by reduced IAP increase, TrA decrease, and spinal compressive force decrease. The increment of C3/C4 shear force affected by the belt also showed an decrease with higher lateral acceleration before the muscle activity exceeded 1. These results show people wearing a belt can bear more lateral acceleration until the muscle is fully recruited.

Table 7. Case study for multi-dimensional vibration on a seated human model

| Case | Vertical Acceleration (G) | Lateral Acceleration (G) |
|---|---|---|
| V (Vertical) | 10 | 0 |
| LA1 (Lateral 1) | 9.98 | 0.5 |
| LA2 | 9.95 | 1 |
| LA3 | 9.88 | 1.5 |
| LA4 | 9.8 | 2 |
| LA5 | 9.54 | 3 |
| LA6 | 8.66 | 5 |




(A) (B)

(C) (D)

Figure 12. Belt effects on multi-dimensional acceleration for a seated human model. (A) intra-abdominal pressure, (B) transverse abdominis activity, (C) L4/L5 compressive force, (D) C3/C4 cervical shear force. The red dots indicate muscle activity higher than 1, which is induced by high lateral accelerations.

## 4. Discussion

### 4.1. Belt effect on lumbar region loading and muscle activity

For all of the shock conditions tested, the abdominal belt reduced the spinal loading in the lumbar region. This decrease in loading was due to the load being partially supported by the IAP. In no case did the belt make the response to the shock more severe in the lumbar region, thereby indicating a protective effect.

When the belt covers the entire lumbar region, all the transverse abdominis have the same muscle activity. However, a reduction of belt width might lead to increased muscle activity for the muscle that wasn't covered by the belt, and the increment value is inversely proportional to the number of muscles not covered by the belt. When the belt covered four vertebra segments, the activity of the unconstrained muscle increased by 60%, which can be over-recruited. This finding supports that the abdominal belt performed best when it covers the entire lumbar region.

The spinalis muscle belongs to the erector spinae muscle group, assist in extension and flexion of the spine (Jun, Kim Jh Fau - Ahn et al.). Therefore, the muscle activity was significantly lower than transverse abdominis under vertical acceleration as there was nearly zero flexion or extension of the body. Weakened muscle with less strength tended to have less muscle activity; the belt was tested to slightly attenuate the effect. Similar phenomenon was found for strengthened muscles. The increased muscle activity indicated higher actual muscle strength, leading to occurrence of muscle fatigue easier. The effect can be constrained with the help of abdominal belt.

The multi-dimensional acceleration didn't lead to any controversial results for the effects of the belt. For the cases studied, the belt showed a similar effect compared to the vertical-only acceleration. significant reductions of the belt effects were observed for two cases with extremely high lateral acceleration; however, the two cases were associated with measured muscle activity higher than the physical threshold value 1. The muscle is not capable of responding to the lateral shock, thus suggesting a limitation for the lateral acceleration around 3g with the simulation. Beyond the threshold, the modelling indicates non-linear responses and the accuracy of the model is not guaranteed.

When the belt is not applied, people with different anthropometric dimensions have larger absolute values of the intra-abdominal pressure, muscle activity, and spinal compressive force. This might be related to larger abdominal muscle size with larger volume variations of the abdominal cavity. When the belt is applied, the same belt force tends to have better effects for smaller-sized people. This might be caused by the relatively smaller-sized artificial muscle representing the belt can provide less support for the larger-sized body.

**4.2. Belt effect on cervical region loading**

There is a risk that mitigation of loading in one part of the body could have adverse consequences elsewhere due to increased transmission of vibration. Potential hazards could exist in the cervical region due to the stabilisation of the lumbar region. The AnyBody model indicated a slight increase in shear force at C4/C5 (2.35%) due to the belt.

The absolute increment of the shear force in cervical region is approximately 2N. Although these results indicate that the belt only causes small cervical compressive force changes, the potential for increased neck loading should not be eliminated at this stage due to the limitations of the AnyBody model. Furthermore, previous research suggested that WBV can affect head repositioning, i.e., proprioception (Morley and Sikorski 2018). In this case, the increment of shear force could be magnified. The results highlight the importance of further study to inform design and the appropriate balance of risk following the introduction of belts.

### 4.3. Limitations

AnyBody is a modelling tool that is capable of simulating musculoskeletal loading and therefore allows for the analyses reported here. However, it has a limitation such that it does not simulate the dynamic motion of segments and associated geometric non-linearities. Non-linearities of apparent mass and transmissibility under WBV reflected by changes in resonance frequency can be studied with experimental measurements and numerical simulation (Mansfield and Griffin 2000, Coe, Xing et al. 2009, Shabana, Gantoi et al. 2011, Liu and Qiu 2020). Besides, the belt model was limited to simulate only stretchy belts that provided a constant inward force, rather than a body reacting to the presence of a non-stretch belt around the abdomen. The belt is also an idealised cylinder, with no variation in width or rigid design elements that may be present in a physical product. In order to model these, a different approach would be needed, potentially building more complex models combining active elements and finite element analysis (FEA).

This study sought to determine the benefit of an abdominal support for idealised shocks. Since the acceleration of the seat on HSC exists in all three translational and all three rotational axes for real shocks, although the study has included the belt effects for two-axis acceleration, future investigations may wish to consider three-axial vibration data to complete the work.

The current model was scaled to represent male crew. The model was not adapted to represent a female-specific geometry; differences in absolute and relative body dimensions (e.g., pelvic size) could affect both the response to the belt, and the interaction of the belt with the pelvic region. Because an increase in shear force can be harmful to the neck, the belt effect on the cervical region is worthy of further investigation with more advanced biomechanical models.

Subjects involved in the study are male crews from UK Royal Marines, and simulations were made assuming subjects sitting on the high-speed craft. Apart from the abdominal belt, there are other ways to mitigate the spinal pressure or stabilise the abdominal region, such as suspension seats used for force absorption. Therefore, it is evident to say that although the abdominal belt in our model reduces the spinal compressive force, other factors which can contribute to the spinal compressive force reduction, or the increment of the intra-abdominal pressure is worth being studied. However, currently this is out of the scope of the paper, and the spring-and-damping seats could be added into the current model for further study.

## 5. Conclusions

In this paper, a novel biomechanical model was built with AnyBody software to study a potential belt protection for seated humans on HSC. The impact of the belt was evaluated by IAP, muscle activities and spinal compressive forces. Various parameters of sea shock levels (3g-10g), belt width/force and human anthropometric dimensions were discussed. In general, wearing the belt didn't make the response to the shock more severe in the lumbar region for all cases. The results have shown a combined increase in IAP (120%) and a reduction of spinal compressive force by 13% on the risk of injury, indicating a positive belt performance for occupants on HSC under severe sea states. The worst case for this study is a 10g vertical acceleration exerted onroyal marine male crew. With a 12% reduction of spinal compressive force at L4/L5 joint, denfending the results in the study with the fact that the belt can stabilise the lumbar region for subjects under the worst condition. Considering the non-linearity effect of risk, the percentage of reduction of injury could be higher. The risk of injury in the neck region has been estimated to be as low as 2.35% increasing in shear force at the C4/C5 joints. Considering the human being's comfort level, the belt force of 700 N was selected. The belt covering the entire lumbar region showed the best performance. Finally, muscle strength analysis indicates that the belt can either compensate for weakened muscle or constrain muscle recruitment and delay muscle fatigue. This study can provide a preliminary guide to the design of an abdominal belt to protect occupants on the HSC.

## Declaration of Interest Statement

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

## Acknowledgement

The technical aspects of this study were conducted independently and exclusively by Nottingham Trent University with industry liaison provided by BMT Defence and Security UK Ltd. The work was funded through a grant provided by the UK Ministry of Defence (MOD) Defence and Security Accelerator (DASA).

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