Experimental and numerical studies of muscular activations of bracing occupant

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ABSTRACT

Occupants who recognize the approaching crash tend to brace themselves. This reflexive muscular activation can affect the kinematics and kinetics of occupant during the crash event but the mechanisms of potential muscle contraction in car crash event remains poorly understood. A quantitative investigation of muscular activation has been attempted by utilizing dynamometer, sled and EMG devices with human volunteers. The experimental findings have been incorporated into the numerical investigation by utilizing a finite element model of skeletal muscular structure of human body.

Eight male subjects were employed and the maximum amount of voluntary isometric muscular contraction for each limb joint at various joint angles was determined using a dynamometer and surface EMG. To mimic the approaching frontal crash and bracing, each volunteer was asked to brace himself when descending in inclined sled system began. During bracing, steering wheel and pedal forces were measured as well as the EMG signals at the volunteer's shoulder, elbow, wrist, knee and ankle joints. The pressure distributions between volunteer and seat back were also measured using a pressure mat.

Simulation of muscle activation for bracing occupant was performed using an optimization process for the joint muscle force calculations. The musculo-skeletal model with the optimized muscle parameters was utilized to validate its tensing behavior against the experimental results. The computed axial compressive loads on steering wheel were respectively 144N and 178N for two sled heights which correlates quite well with the average value of test measurements (121.7±46.6N and 151.1±78.9N). The computed reaction forces at pedal and seat back also exhibited quite good agreement with the test measurements.

INTRODUCTION

The bracing driver in pre-impact situation tends to extend elbow and knee joints, and consequently push the pelvis back into the seat and lean backward as shown in Fig. 1.

Fig. 1 Postures of driver: Before (left) and after (right) bracing

The bracing induced by reflexive contractions of joint muscles change the kinematics and kinetics of occupant during the crash. Its effects on injury risk have been also investigated: Begeman et al [1] studied the response of the human musculo-skeletal system to impact acceleration. They employed volunteers and EMG technology to identify the muscular response before, during and after the impact acceleration. It was found that the tone of the lower extremity muscles changed the kinematics of occupant and force distribution of restraints. However they only focused on the bracing of the lower extremity and also could not quantify the degree of the muscle activations. Klopp et al [2] also studied the effects of the reflexive bracing, a series of computer simulations, pendulum and sled tests with Hybrid III dummies and human cadavers. It was concluded that the effect of muscular preloading was to increase the efficiency of load transmission to the leg and the preloaded legs acted as additional restraints helping the occupant ride down the crash pulse. Gordon et al [3] performed static and dynamic characterizations of human hip, knee and ankle. They computed forces and torques acting on the joints by measuring seat and pedal loads.

In this study, the muscular activation of bracing occupants was quantified using a dynamometer, sled system and EMG devices with human volunteers. A deliberate process was taken in the selection of volunteers since the individual divergence in muscular structure between the volunteers might generate large deviations in the bracing test. Therefore, total 8 volunteers having similar body compositions as well as anthropometries were selected. Using the dynamometer, isometric voluntary maximal torques for 5 joints, shoulder, elbow, wrist, knee, and ankle of each volunteer were characterized. EMG signals at the pair of muscles, each representing an extensor and a flexor were also monitored for various joint angles. Assuming the maximal voluntary effort was made, the extension and flexion should have brought the maximum levels of EMG signals from the associated muscle group. The mean rectified EMG signals from the maximally contracted muscle were utilized as a reference value for computing the activation level of corresponding muscle in bracing test with a sled system. To mimic the approaching frontal crash, the inclined sled system driven by gravity was designed and built as shown in Fig. 2. Each volunteer was asked to brace himself when descending began on the slope until the sled stopped by striking an energy-absorbing barrier. During the bracing in the sled, steering wheel and pedal forces were measured from the installed load cells as well as the EMG signals from the volunteer's joint muscles. The pressure distributions between the back of volunteer and seat were also measured using a pressure mat

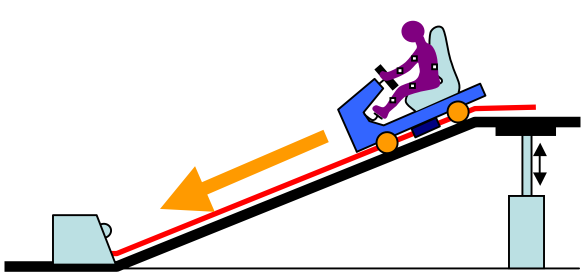


Fig. 2 Sled system for bracing test

Finite element modeling of skeletal muscular structure and numerical investigation of its activations were performed subsequently to the experimental study. An optimization scheme based on an ergonomic criterion [4] was adopted for the calculation of internal muscular force distributions around joints. The muscle tensing behavior of the model was validated against the test results.

EXPERIMENTAL STUDY

Selection procedure of volunteers

The lean balance, ratio of muscle mass to the body weight and isometric voluntary maximum torques at elbow and knee joints were extra indices in addition to the anthropometric data for selecting volunteers. During the first round of the two-stage selection process, 20 out of 128 volunteers were selected based on BMI (Body Mass Index, kg/m2). The selection criterion of the BMI was 22±1 kg/m^2 (Height: 1.75±0.01m, Body mass: 67±1kg). Isometric voluntary maximal elbow and knee joint torques had been measured with those 20 volunteers in the second round and 8 volunteers with responses closest to mean values were then chosen for the final tests. As a consequence, the dispersion in the final group of volunteers, e.g., standard deviation of maximal joint torques had decreased from the first round selections by 41% and 26% for elbow and knee joints, respectively. The average and standard deviation of the final 8 volunteer's anthropometric data and body compositions are listed Table 1

Table 1 Volunteer data

	Age	Height (m)	Weight (kg)	BMI (kg/m²)	RA lean balance (%)*	RL lean balance (%)**
Mean	24.2	1.746	67.31	22.09	4.62	13.28
S.D	1.69	0.84	1.55	0.61	0.31	0.58

*: Ratio of right arm muscle mass to total body mass (%)
**: Ratio of right leg muscle mass to total body mass (%)

Measurement of isometric maximal joint torque and voluntary muscle contraction using dynamometer

In order to gauge the maximal voluntary contractions (MVC) of selected muscles, each volunteer was asked to produce the utmost isometric joint torques in a dynamometer (model: Biodex™ System 3 Pro). The EMG activities of a pair of muscles for extension and flexion were simultaneously measured using surface electrodes. The selected joint muscles for EMG measurement are listed in Table 2.

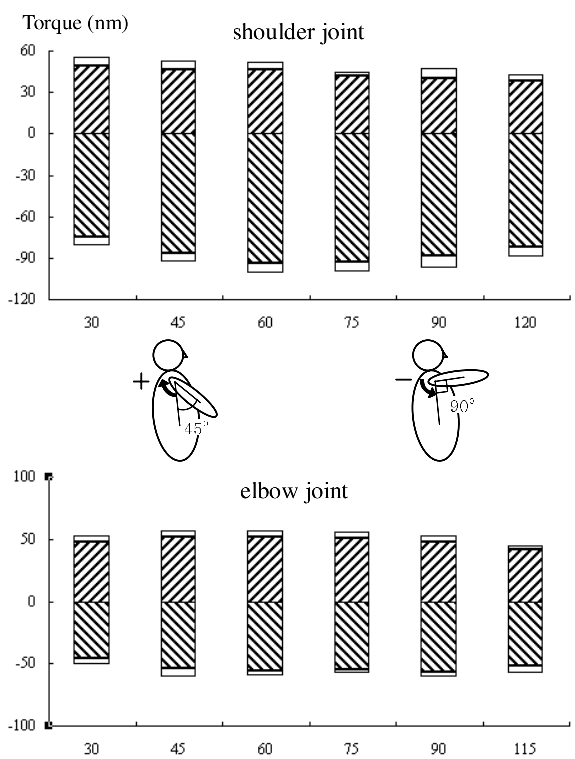
Table 2 Joint muscles for EMG activity monitoring in dynamometer test

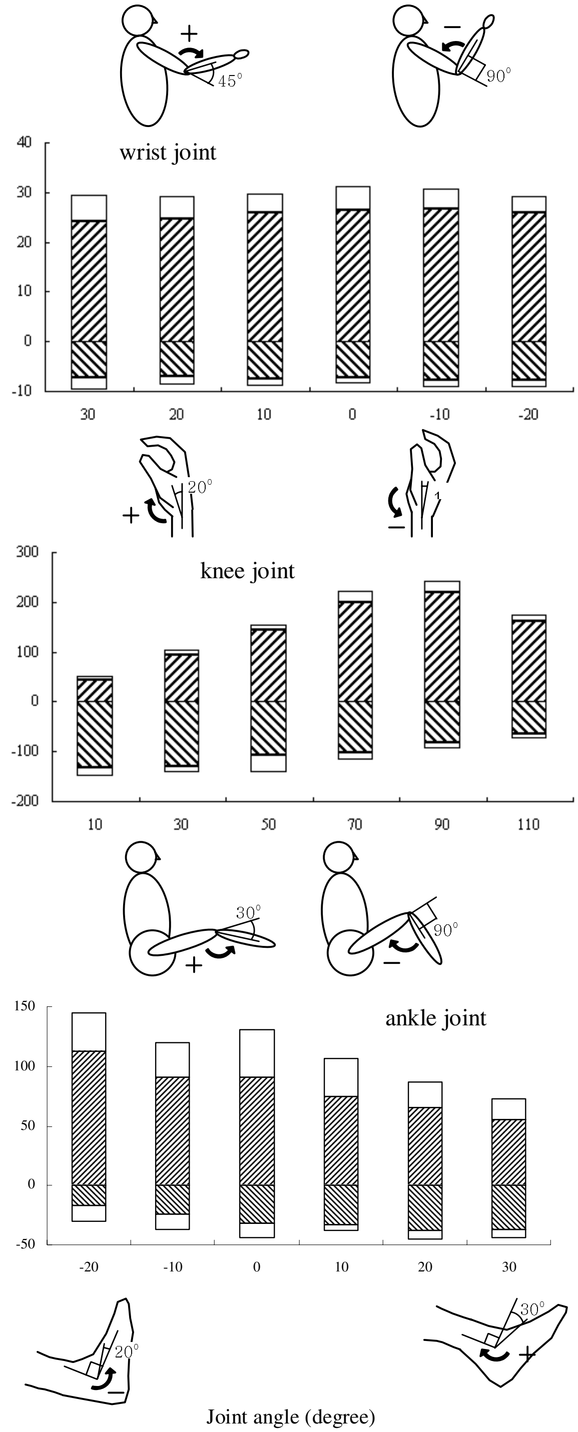
Muscle / Joint	Extensor	Flexor
Shoulder	Posterior deltoid	Anterior deltoid
Elbow	Medial triceps	Biceps brachii
Wrist	Extensor capri radialis	Flexor capri radialis
Knee	Rectus femoris	Biceps femoris
Ankle	Soleus*	Tibialis anterior**

*: plantaflexor, **: dorsiflexor

The dynamometer test setup with a volunteer is shown in Fig. 3. The measured maximal joint torques with various joint angles for five joints in upper and lower limbs (shoulder, elbow, wrist, knee, ankle) during isometric muscle contractions are shown in Fig. 4.

Fig. 3 Measurement of maximal voluntary joint torque in a dynamometer





Joint angle (degree)

(◩: extension(+), ◪: flexion(-), □: S.D.)

Fig. 4 Maximal isometric voluntary joint torques from dynamometer test

During the dynamometer test, EMG activities of representative pairs of joint muscles in Table 2 were monitored. Fig. 5 shows a typical raw data set of the elbow joint composed of torque and EMG signals obtained from dynamometer and surface electrodes, respectively.

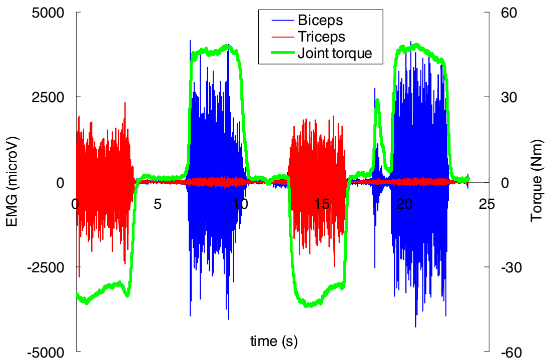
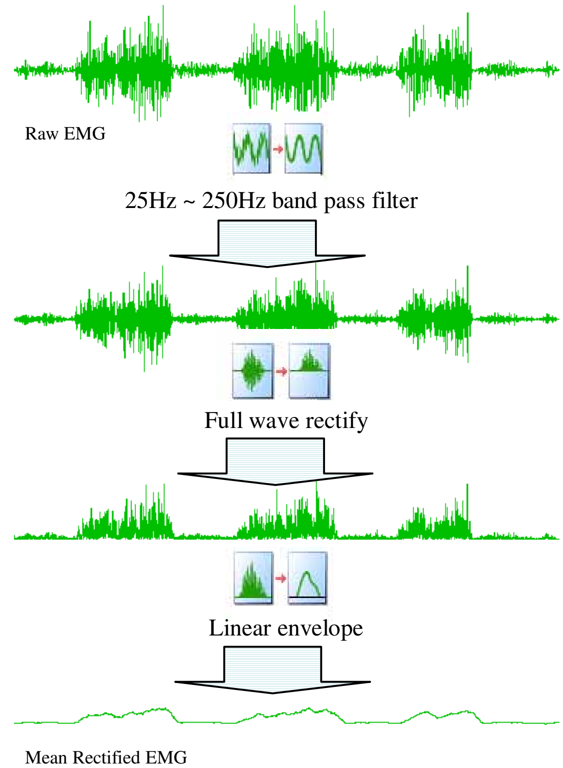


Fig. 5 Typical raw EMG signals and joint torque from dynamometer test (Elbow joint at 75^{o} angle)

Processing the raw EMG signal, rectifying, filtering (low pass filter: 250 Hz, high pass filter: 25Hz), and smoothing (LP filtering), a MR EMG (Mean Rectified EMG) signal was obtained as shown in Fig. 6. A RMS (Root Mean Square) value was then computed from the MR EMG signal, which represents an intensity of the EMG signal and an index of muscle activation level at the maximal voluntary contraction (MVC). There were considerable divergences in RMS values between volunteers in spite of their similar lean balances. This might be due to the different amount of subcutaneous fat tissue between volunteers and variability in electrode positioning relative to active muscle fibers.



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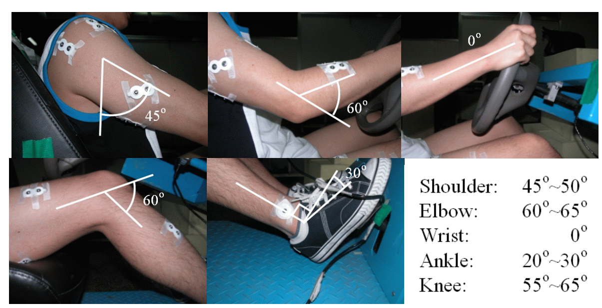
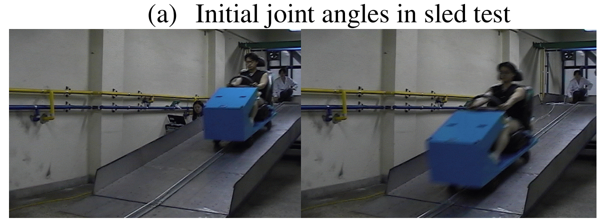
ig. 6 EMG signal processing

Measurement of activation level of bracing muscles in sled system

Fig. 7 shows a volunteer's configuration in sled test. Two different sled heights, 0.9m and 1.0m were tried twice each and the measurements were very repeatable.



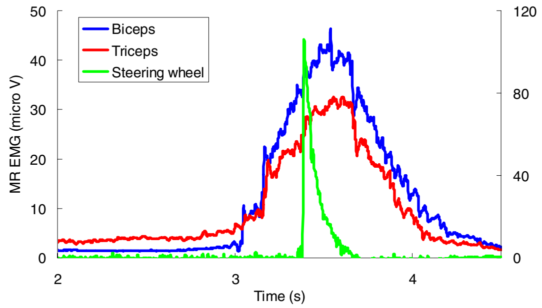
(a) Initial joint angles in sled test



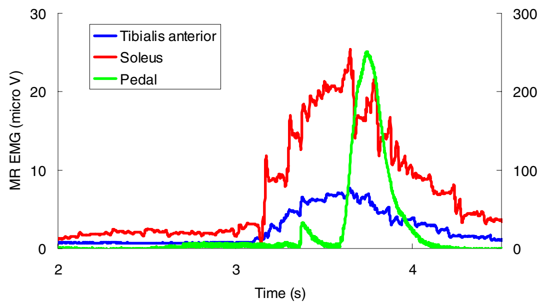
(b) Volunteer in descending slope
(Left: before bracing, Right: after bracing)

Fig. 7 Configuration of volunteer in sled test

Typical profiles of reaction forces at steering wheel and pedal are presented in Fig. 8 with mean rectified EMG signals monitored at the muscles of the elbow and ankle joints. In general, the reaction force developed 0.3-0.5s after the onset of EMG activity, which is similar to the timing observed in an earlier volunteer test [1].



(a) Upper limb



(a) Lower limb

Fig. 8 Reaction force profiles with muscle activities (MR EMG)

The volunteer's pattern of bracing in sled test was quantified by computing his ratio of joint muscle activation levels to the maximal voluntary contractions from dynamometer test mentioned in previous section. Fig. 9 shows the %MVCs, the ratios of RMS of MR EMG signals between sled and dynamometer tests. The higher sled at 1.0m height tends to induce from 5% to 20% more muscle activations in both extensors and flexors than the lower sled at 0.9m height, -except the knee joint. But quite same ratios of activations between extensors and flexor were produced from both sled heights. The extensors were significantly more activated than flexors in elbow, wrist and ankle joints while the opposite tendency found in shoulder and knee joints. There were relatively large standard deviations in the sled test comparing to the dynamometer test since the styles of the bracing might have differed greatly between volunteers.

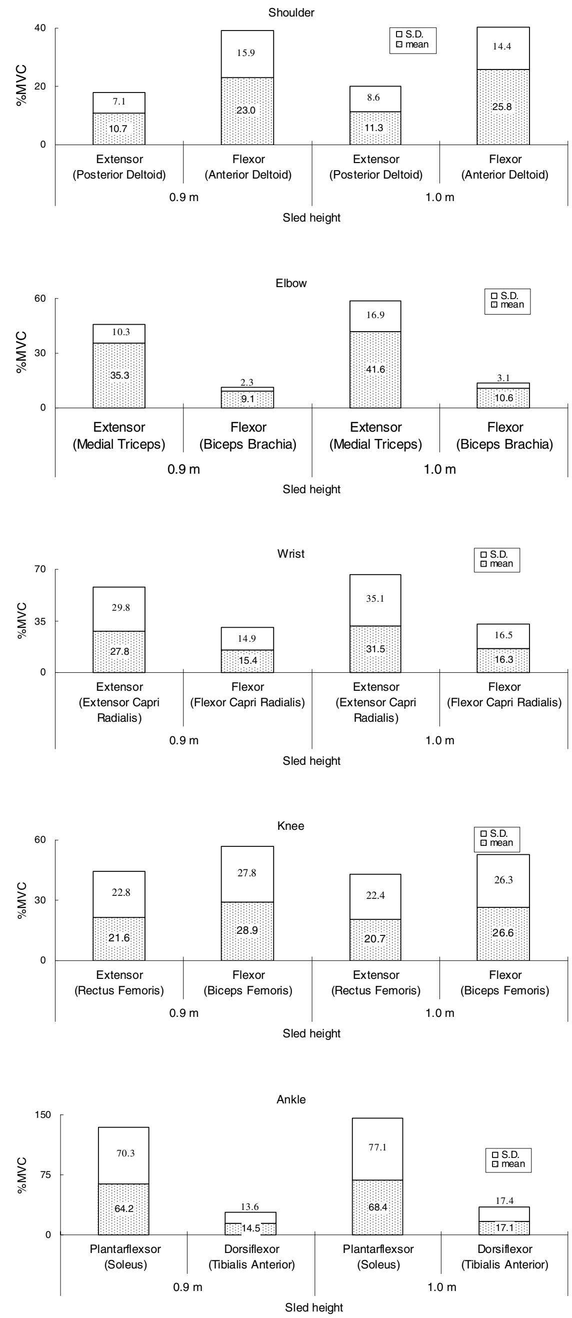


Fig. 9 %MVC of joint muscles from sled test

The average axial forces measured at steering wheel and pedal as shown in Fig. 10 also indicate that volunteers braced more at the higher sled drop resulting in larger reaction forces.

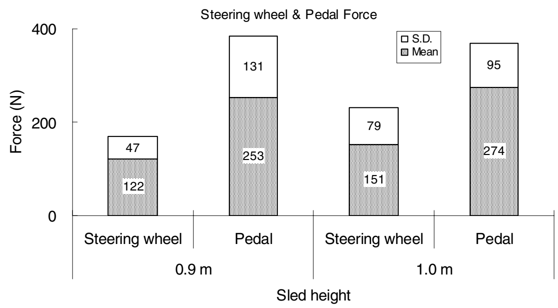


Fig. 10 Reaction forces at steering wheel and pedal

Elbow joint extension and subsequent rearward rotation of the upper body developed more contact pressure on seat back as displayed in Fig. 11. The average net normal reaction force at seat back, the area integration of increased pressure by bracing is shown in Fig. 12.

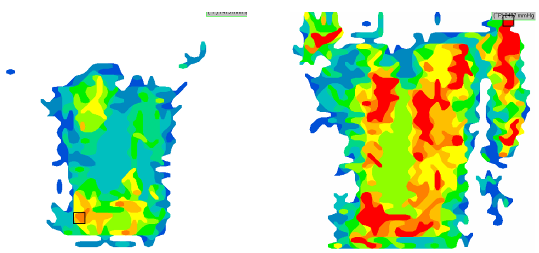


Fig. 11 Measured contact pressure distributions between volunteer and seat back

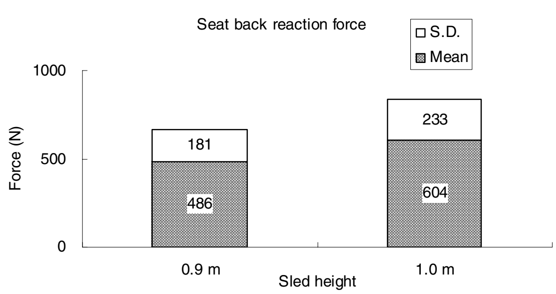


Fig. 12 Reaction force at seat back

NUMERICAL STUDY

Numerical investigation of muscular activation was performed subsequent to the experimental study presented in the first part of this paper.

Human body modeling

The H-model, shown in Fig. 13, is a finite element human body model representing the 50% male anthropometry. This model is widely used for crashworthiness simulation [5]. Each body segment in the H-model, in a version aiming for muscle tensing simulation, was defined as a rigid body and was linked by the anatomical joints and with the relevant skeletal muscles represented by bar finite elements [5]. The incorporated sixteen major skeletal muscles modeled by Hill type one dimensional bar elements [6] are listed in Table 3. The articulated joints were modeled with kinematic joint elements whose characteristics were designed to have no resistance within the range of motion such that only muscle forces could be the source of joint torques. Seeking the average of active isometric muscle force-length relations of the model, the maximal forces (F_{max}) of each muscle at various lengths with different joint angles were computed based on the maximal isometric voluntary joint torques obtained from dynamometer test in Fig. 4. In the case when multiple muscles were involved for the same articulation DOF, e.g., biceps brachii, brachialis, and brachioradialis for elbow flexion, an optimization algorithm was adopted to determine the likely distribution of the muscle forces (design variable) by minimizing the active muscle energy (objective function) for static equilibrium (constraints). The sequential response surface method in HyperOpt [7] was selected for the optimization process. Fig. 14 shows the result of computed isometric maximum muscle forces for shoulder, elbow, knee, and ankle joints.

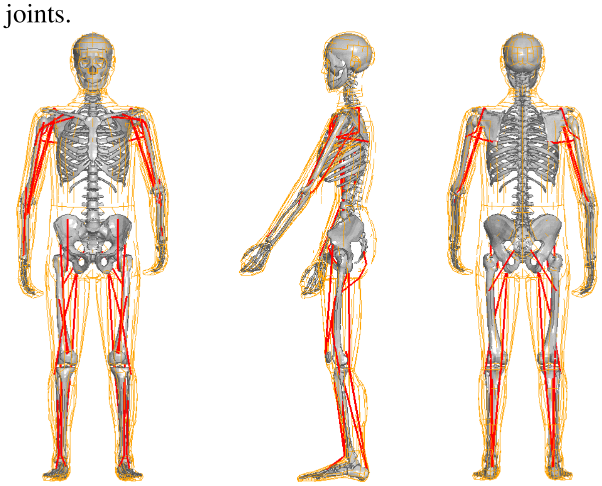
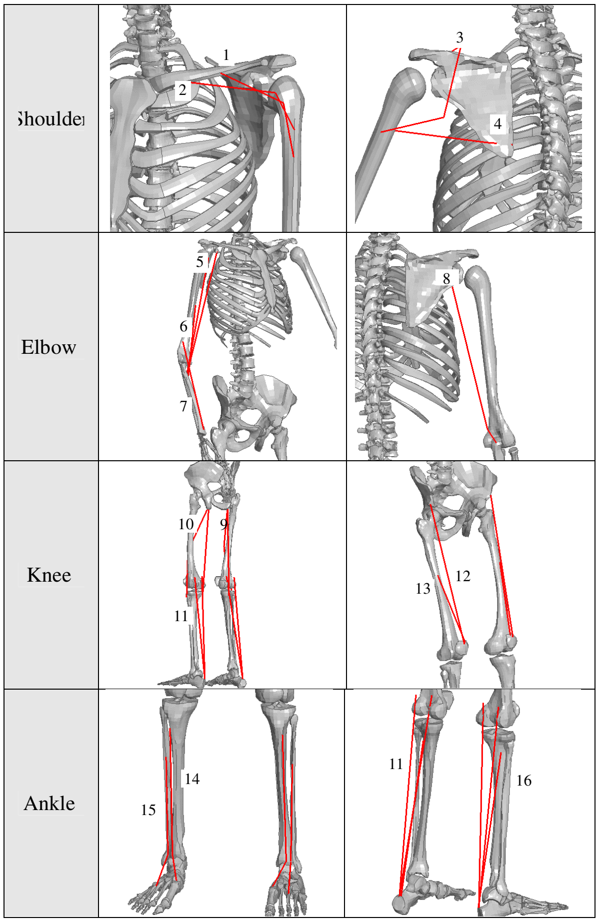


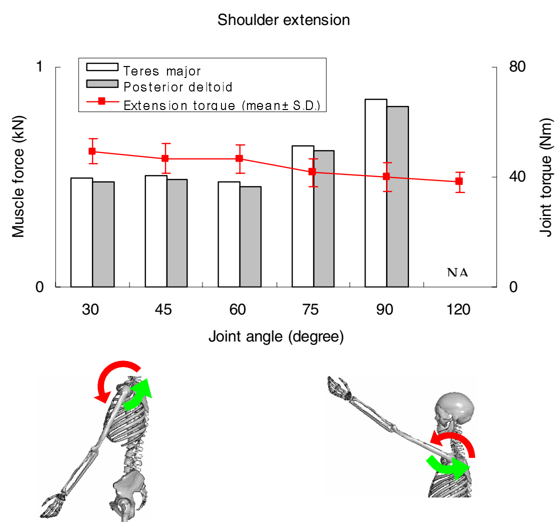
Fig. 13 H-model with skeleton and muscles

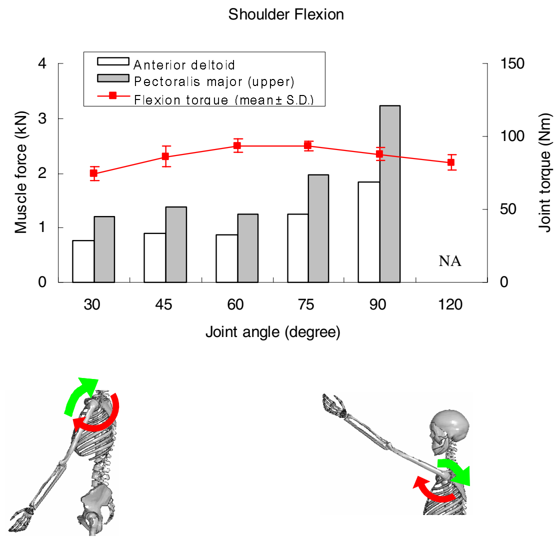
Table 3 Skeletal muscles in H-model for the simulation of bracing

	Flexors	Extensors

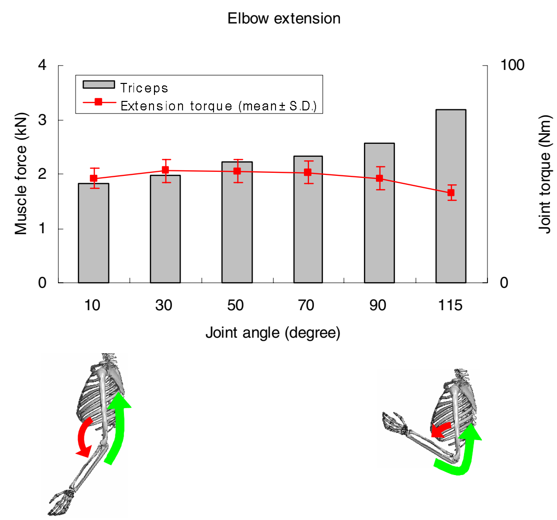


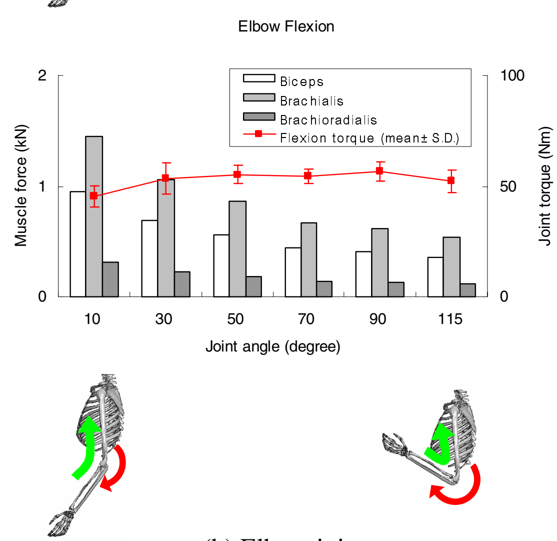
1. Anterior deltoid
2. Pectoralis major (upper part)
3. Posterior deltoid
4. Teres major
5. Biceps
6. Brachialis
7. Brachioradialis
8. Triceps
9. Biceps femoris
10. Semitendonous
11. Gastrocnemius (lateral & medial)
12. Rectus femoris
13. Vastus intermedius
14. Tibialis anterior
15. Extensor digitorium
16. Soleus



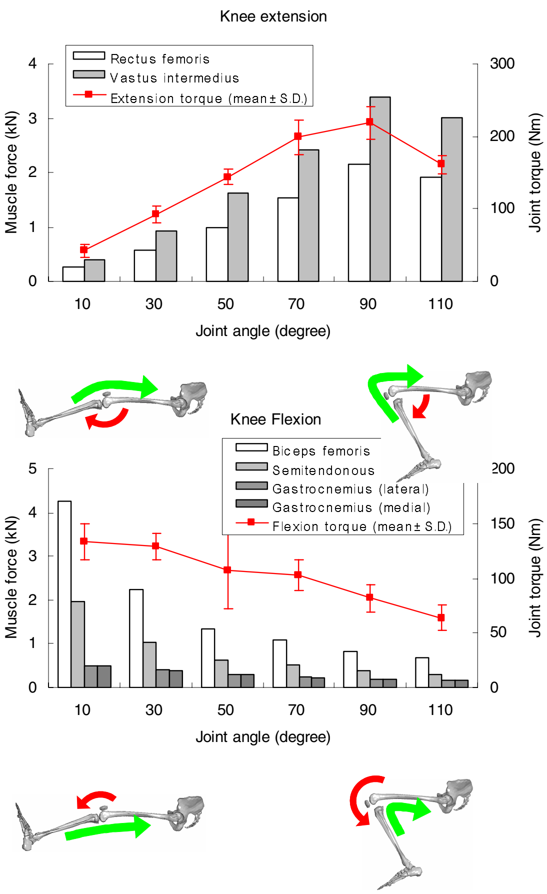


(a) Shoulder joint

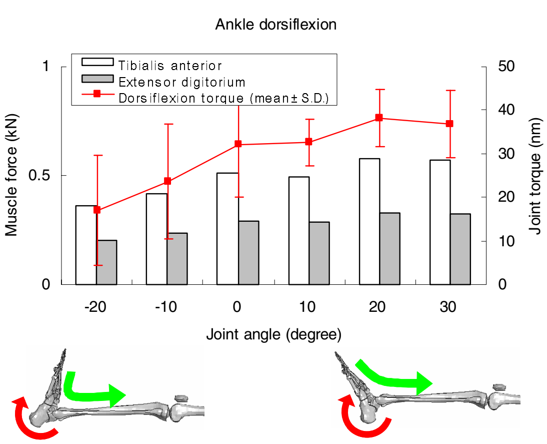


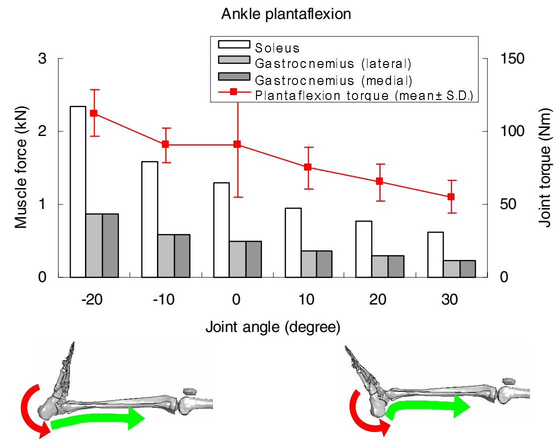


(b) Elbow joint



(c) Knee joint





(d) Ankle joint

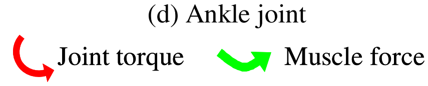
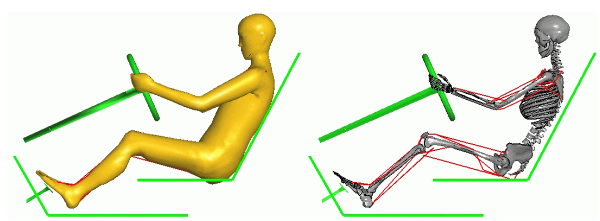


Fig. 14 Computed isometric maximum muscles forces (F_{max}) (wrist joint was not performed)

Simulation of bracing occupant

The seat, floor panel and steering wheel of the sled system were added to the H-model with driving posture as shown in Fig. 15. Sliding contact interfaces were defined between the seat and the skin part of the H-model. Translational motions of hands and feet were respectively tied to steering wheel and pedal such that the forces generated from the muscle bracing could be transmitted.



Displayed with skin   without skin

Fig. 15 Configuration of H-model for the simulation of bracing occupant

In the simulation, the average values of %MVC in joint muscles, the ratios of RMS of mean rectified EMG signals between sled and dynamometer tests which are listed in Table 4 were applied as activation levels of the bracing muscles. The reaction forces at the steering wheel, pedal, and seat back were then computed until they statically equilibrated with the imposed bracing muscle forces. The simulation results correlate quite well with the experimental measurements as shown in Fig. 16

Table 4 Average volunteer's muscle activation levels used for bracing simulation

Sled height	0.9 m			1.0 m		
Muscle / Joint	xtensc	Flexoı	Ratio*	xtensc	Flexoı	Ratio*
Shoulder	10.7	23.0	0.46	11.3	25.8	0.44
Elbow	35.3	9.1	3.88	41.6	10.6	3.93
Wrist**	15.4	27.8	0.55	16.3	31.5	0.52
Knee	21.6	28.9	0.75	20.7	26.6	0.78
Ankle	64.2	14.5	4.43	68.4	17.1	3.99

*: Ratio=Extensor/ Flexor, ** Wrist joint is not included in the model

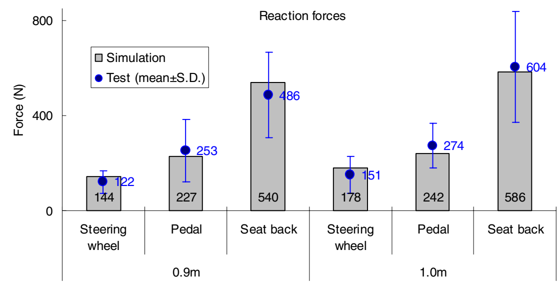
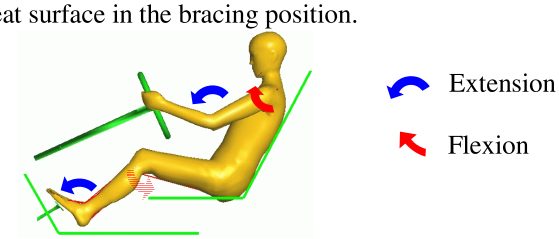
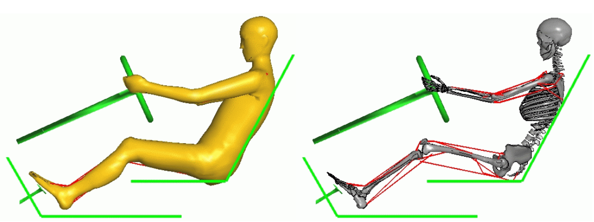


Fig. 16 Comparison of reaction forces between volunteer test and simulation

The simulated driving posture altered by muscle tensing is illustrated in Fig. 17. There is a noticeable straightening of arms and an elastic penetration into the seat surface in the bracing position.



Initial posture and joint torques (before bracing)



After bracing

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Fig. 17 Simulated bracing posture

The computed peak muscle forces during the bracing, which are proportional to the activation levels multiplied by the isometric maximum voluntary forces at corresponding joint angles, are shown in Fig. 18. The net joint torque generated by tensing of each muscle depends on the effective moment arm of the muscle with respect to the corresponding joint center.

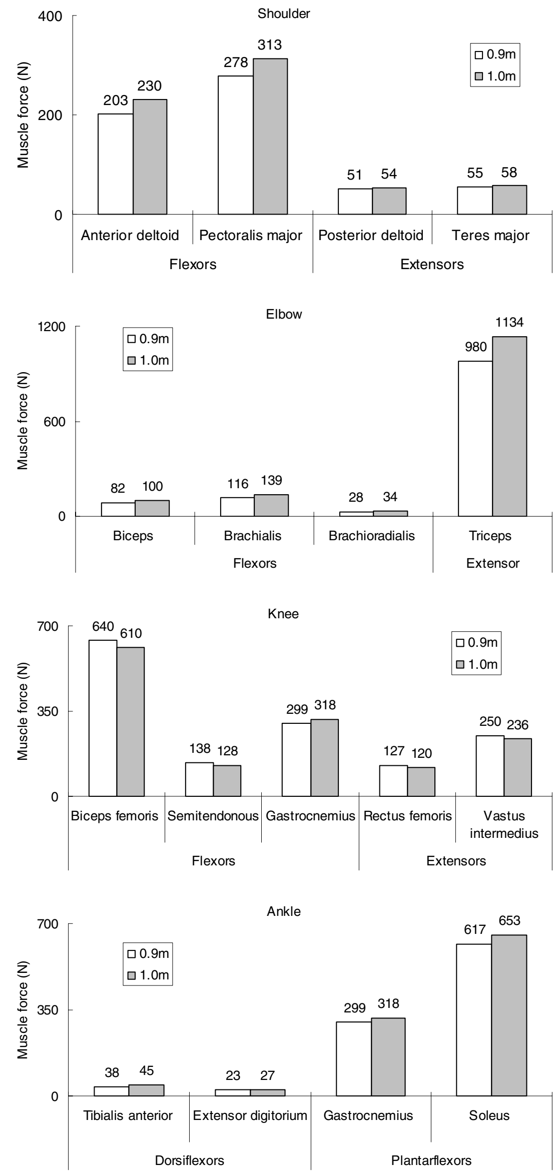

Fig. 18 Computed bracing muscle forces

CONCLUSIONS

Vehicle occupants tend to brace in anticipation of a crash and this pre-crash muscle tensing can change the kinematics and kinetics of the occupants. The pattern of extremity bracing, i.e., shoulder, elbow, wrist, knee and ankle joints was quantitatively analyzed by volunteers EMG test. For shoulder, elbow and ankle joints, activations of extensors were substantially higher than those of flexors. However, an opposite trend was found at wrist and knee joints. The reaction forces at steering wheel, pedal and seat back were also measured to identify the degree of muscle tensing.

Numerical simulation of muscle tensing was performed to verify the finite element human body model. The simulated muscle tensing behavior of the model such as amounts of reaction forces at the steering wheel, pedal and seat back correlated quite well with the test results. It was the first step in the development of human body model to investigate the effect of muscle tensing on occupant kinematics and kinetics. A crash simulation with likely dynamic muscle activations taken into consideration would follow as a next step.

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