

A NEW LEGFORM IMPACTOR FOR EVALUATION OF CAR AGGRESSIVENESS IN CAR-PEDESTRIAN ACCIDENTS

Adam Wittek, Atsuhiko Konosu, Yasuhiro Matsui, Hirotoshi Ishikawa

Japan Automobile Research Institute

Akira Sasaki

Japan Automobile Manufacturers Association

Japan

Tariq Shams, Jason McDonald

GESAC Inc.

United States

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ABSTRACT

The goal of the present study was to develop a new legform impactor that accurately represents both the impact force (i.e., force between the leg and impacting mass) and leg kinematics in lateral impacts simulating car-pedestrian accidents. In its development we utilized the knee joint of the pedestrian dummy called Polar-2 (HONDA R&D) in which the cruciate and collateral ligaments are represented by means of springs and cables, the geometry of the femoral condyles is simplified using ellipsoidal surfaces, and the tibial meniscus is represented by an elastomeric pad.

The impactor was evaluated by comparing its responses with published experimental results obtained using postmortem human subjects (PMHS). The evaluation was done under two conditions: 1) impact point near the ankle area (bending tests), and 2) impact point 84 mm below the knee joint centre (shearing tests). Two impact speeds were used: 5.56 m/s and 11.11 m/s.

The responses of our impactor were reasonably close to those observed in the experiments using PMHS in terms of both impact force and leg shearing displacement (i.e., relative displacement between the leg and thigh at the knee joint level in a lateral direction). In the shearing tests, the peak values of leg shearing displacement were greater than 30 mm.

INTRODUCTION

In non-fatal car-pedestrian accidents, lower extremities account for around 40% of the most commonly injured body parts (ITARDA, 1996). These injuries often lead to long-term or permanent disability, and their reduction is one of the priority items in traffic safety strategy. An important element of such strategy is to decrease the aggressiveness of the components of a car front. A commonly used method of evaluation of such aggressiveness is subsystem testing using legform impactors. Several such impactors have been developed so far. The most widely utilized are those of the Transport Research Laboratory (TRL impactor)

(EEVC, 1998) and the Japan Automobile Research Institute (JARI-1 impactor) (Matsui et al., 1999). The TRL impactor has already been used in the European New Car Assessment Program (Euro NCAP) and has been accepted as a prototype test device in a working draft of the European Regulations (EEVC, 1998). However, the recent study by Matsui et al. (1999) has indicated important differences between the responses of both TRL and JARI-1 impactors and the behavior of human lower extremities. According to this study, these impactors exhibit much lower peak values of relative displacement between the leg and thigh at the knee joint level in a lateral direction (i.e., leg shearing displacement) than those observed in experiments conducted on postmortem human subjects (PMHS). This leads to a question about the validity of the TRL and JARI-1 impactors in evaluations of car front aggressiveness, since the leg shearing displacement is often used as an indicator of injury risk to the knee joint ligaments.

Therefore, the goals of the present study are as follows. First, to develop a new legform impactor (referred to as the new JARI impactor) that accurately represents both the impact force and leg kinematics in lateral impacts simulating car-pedestrian accidents. Second, to evaluate the biofidelity of this newly developed impactor by comparison of its responses with the results of PMHS experiments. Third, to compare biofidelity of the TRL and new JARI impactors.

METHODS

Development of New JARI Legform Impactor

Knee Joint It is likely that the lack of biofidelity of the TRL and JARI-1 impactors reported by Matsui et al. (1999) results from the too simplified structure and geometry of knee joints of these impactors. None of them represents the knee articular surfaces, and in both of them the ligaments are simplified by means of metal bars. In consequence, in both TRL and JARI-1 impactors, the relative displacement between the leg

and thigh in a lateral direction is too strongly constrained.

For this reason, in the new JARI impactor, the human knee joint structure and geometry were more accurately represented. This was done by application of the knee joint of the pedestrian dummy POLAR-2 developed by GESAC and HONDA R&D (Artis et al., 2000). In this joint, the femoral condyles are simplified by means of ellipsoidal surfaces with a left/right symmetry, and the tibial meniscus is represented by means of a urethane pad (Figures 1 and 2a). To assure the durability of this pad, its thickness was arranged to exceed that of the human meniscus. For the same reason, the intercondylar eminence was made broader than its human counterpart. In contrast to the TRL and JARI-1 impactors, the knee joint ligaments in the new JARI impactor are represented by cables connected to a system of non-linear springs and rubber tubes (Figures 1 and 2a). The bending stiffness of these cables is very low, and they can constrain the leg motion only through their tensile forces. These forces are determined by the damping and force-elongation properties of the system of springs and rubber tubes as described by Artis et al. (2000). Another important feature of the spring and cable representation of the ligaments is that these elements are reusable in contrast

to the metal bars in the TRL and JARI-1 leg impactors which have to be replaced after a test.

Leg and Thigh As with the TRL and JARI-1 impactors, the leg and thigh shafts in the new JARI impactor are made of very stiff aluminum tubes. They were designed according to the geometrical and mass/inertia specifications indicated in a draft proposal of the ISO standard (ISO, 1996) (Figure 2).

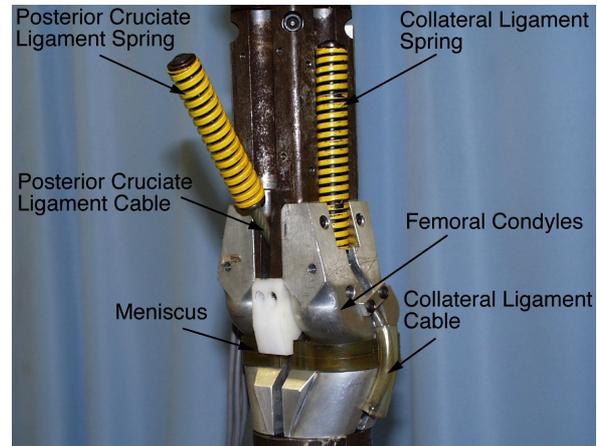
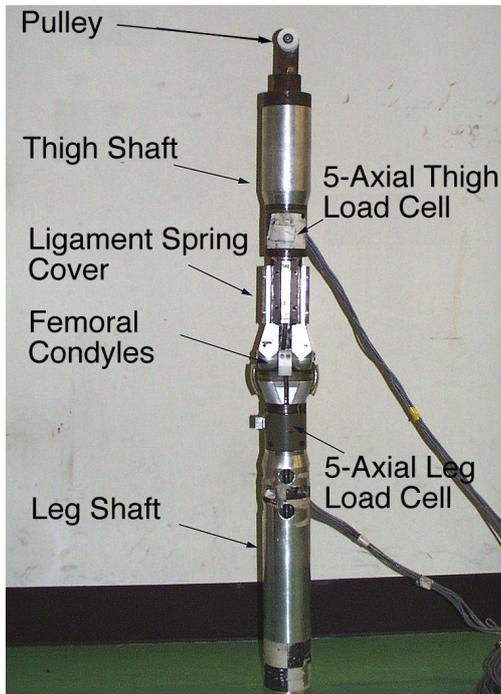


Figure 1. A front oblique view of the knee joint of the new JARI impactor.

a)



b)

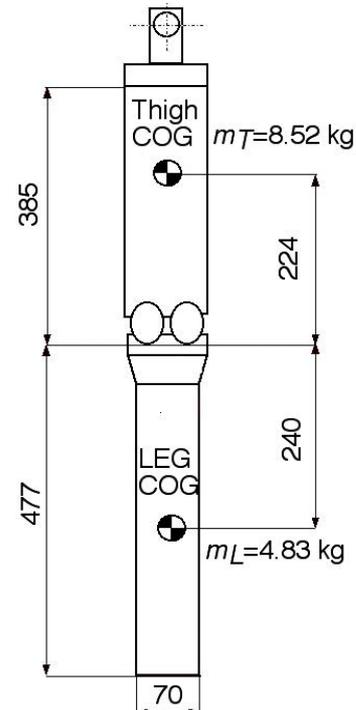


Figure 2. The new JARI legform impactor: a) Front view; b) Position of gravity centers (COG) of the impactor leg and thigh. m_L and m_T are masses of the impactor leg and thigh, respectively. m_L and m_T include the impactor foam and “skin” (for explanation see the section *Experimental Set-Up and Test Matrix*). Dimensions are in millimeters.

Evaluation of Biofidelity of New JARI Legform Impactor

Experimental Set-Up and Test Matrix The new legform impactor was evaluated against the responses of PMHS legs determined by Wittek et al. (2000) using the results of the experiments conducted by Kajzer et al. (1997, 1999). In these experiments, the PMHS legs were impacted in a lateral direction at two speeds: 1) 5.56 m/s (20 km/h) and 2) 11.11 m/s (40 km/h). Two impact configurations were used: 1) shearing (impact point at the fibula head level) (Figure 3), and 2) bending (impact in the ankle area) (Figure 4).

In the present study, we conducted eight impact tests of the new JARI impactor under conditions closely replicating the set-up of the experiments by Kajzer et al. (1997, 1999) (Figures 3 and 4, Table 1). Duplicating their experimental procedure, we struck the impactor leg with a metal ram padded with one layer of Styrodure™ foam of the same dimensions as those used in the PMHS experiments by Kajzer et al. (1997, 1999). The mass of our ram differed by only 0.02 kg from theirs.

To reproduce the 400 N pre-load applied to the PMHS by Kajzer et al. (1997, 1999), we pre-loaded the impactor with a 41.6 kg mass attached to the impactor top by means of a spherical joint (Figures 3 and 4). To

simulate the constraints applied to the PMHS thighs, we supported the impactor thigh with two bolts of the same diameter as that used by Kajzer et al. (1997, 1999). To represent the ground, we used thick steel and teflon plates in the shearing and bending tests, respectively. The teflon plate was applied to minimize the effects of friction between the impactor leg and the ground in the bending tests.

As already mentioned, the leg of the new JARI impactor consists of a very stiff cylindrical shaft. The contact stiffness between such a shaft and the foam padding of the side ram is likely to be higher than that of the contact between the padding and a human leg. Therefore, in all our tests, a 25 mm layer of memory foam (Confor™ foam by Ear Specialty Composites Corp., USA) was attached to the impactor leg to represent a human leg flesh. The use of this foam is recommended in a working draft of the European Regulations (EEVC, 1998).

In the experiments by Kajzer et al. (1997, 1999), the position of the impact point varied because of differences in the size of the PMHS bodies. Therefore, when evaluating the new JARI impactor, we used the average position of the impact point determined in these experiments: 84 mm and 377 mm below the knee joint center in the shearing and bending tests, respectively.

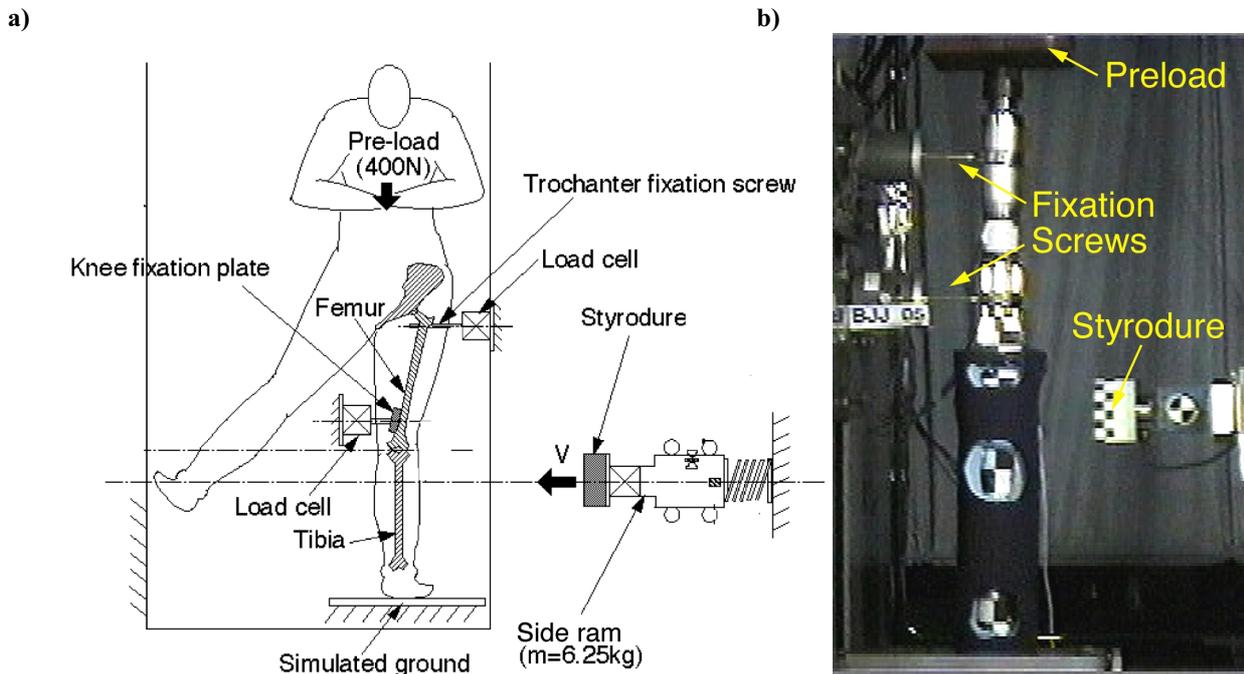


Figure 3. a) Set-up of the shearing-type PMHS experiments by Kajzer et al. (1997, 1999). Based on Kajzer et al. (1997). b) Set-up of the shearing-type biofidelity tests of the new JARI legform impactor.

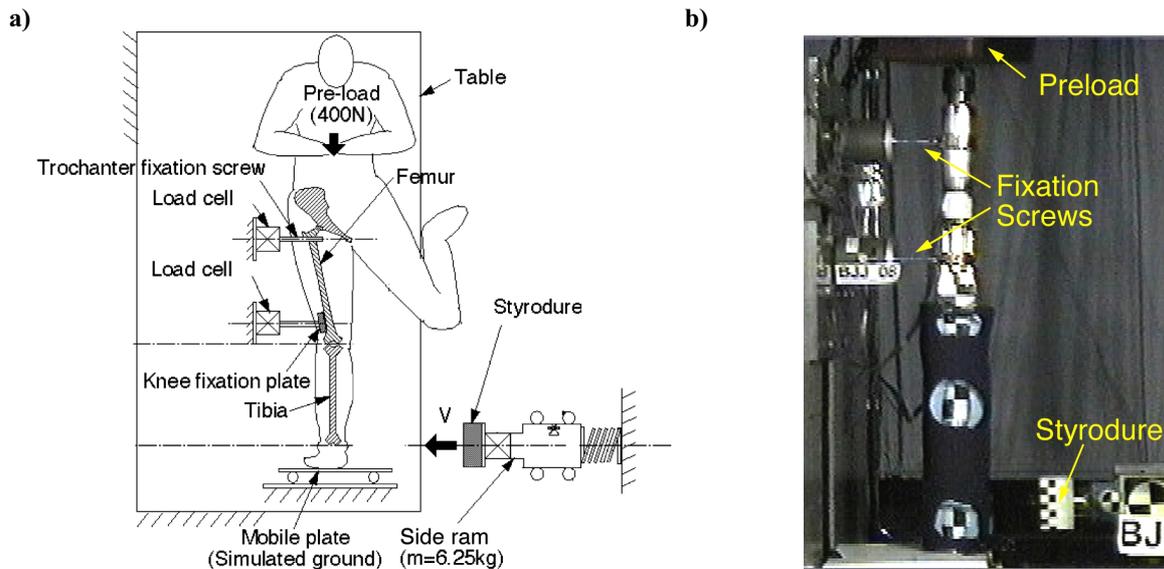


Figure 4. a) Set-up of the bending-type PMHS experiments by Kajzer et al. (1997, 1999). Based on Kajzer et al. (1997). b) Set-up of the bending-type biofidelity tests of the new JARI legform impactor.

Table 1.
Test matrix for evaluating biofidelity of the new JARI impactor

Experiment Number	Type of Impact	Impact Speed m/s (km/h)
BJJ02, BJJ03	Shearing	5.56 (20.0)
BJJ04, BJJ05	Shearing	11.11 (40.0)
BJJ06, BJJ07	Bending	5.56 (20.0)
BJJ08, BJJ09	Bending	11.11 (40.0)

Analyzed Variables In evaluation of the biofidelity of the present impactor, we compared its impact force, leg shearing displacement and leg bending angle with the corridors (average \pm standard deviation SD) of the responses of the human lower extremity determined by Wittek et al. (2000) using the results of PMHS experiments by Kajzer et al. (1997, 1999). Following Kajzer et al. (1997, 1999), we calculated the impact force (i.e., the force between the impactor leg and the impacting side ram) as the product of the ram mass and acceleration. The leg shearing displacement D was determined by means of the following formula:

$$D = X_{P4} - X_{P3} - g \sin(\alpha), \quad (1)$$

where g is the position of marker P4 in relation to the center of the knee joint measured along the longitudinal leg axis, and α is the bending angle of a leg. X_{P3} and X_{P4} are coordinates of markers P3 and P4 in a lateral direction, respectively (Figure 5). The leg bending angle α was obtained as follows:

$$\alpha = \arctan\left(\frac{X_{P2} - X_{P1}}{Z_{P1} - Z_{P2}}\right), \quad (2)$$

where X_{P1} and X_{P2} are coordinates of markers P1 and P2 in a lateral direction, and Z_{P1} and Z_{P2} are coordinates of markers P1 and P2 in a vertical direction.

Coordinates of all the markers were determined from high-speed video tape digitized and analyzed by means of the NAC Image-Express workstation (NAC, 1995). The tape recording speed was 500 frames per second. This implies that when calculating the leg bending angle and leg shearing displacement, we were able to determine the start of impact (i.e., zero on the time axis) with an accuracy not greater than 2 ms.

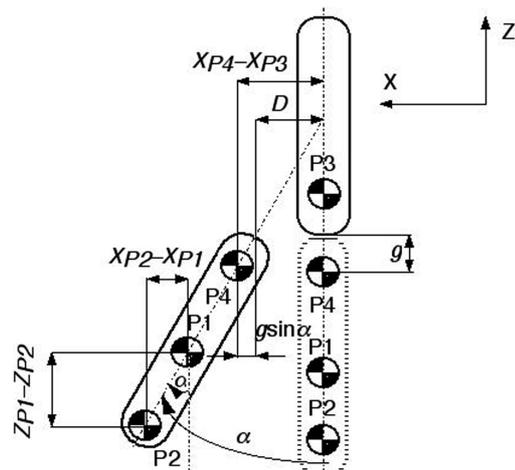


Figure 5. Scheme of calculation of leg shearing displacement D and bending angle α .

Comparison of Biofidelity of TRL and New JARI Impactors

The responses of the TRL impactor used here were from Matsui et al. (1999) who conducted their experiments without a damper attached to the impactor shear system. These responses could be compared with the behavior of the new JARI impactor to a limited extent only because the TRL impactor was not evaluated in shearing-type impacts at a speed of 11.11 m/s to minimize the risk of damage.

RESULTS

Impact Force

In shearing-type tests at an impact speed of 11.11 m/s and for bending-type tests at both impact speeds of 5.56 m/s and 11.11 m/s, the peak values of the impact force-time histories of the new JARI impactor were within the response corridors determined using the results of the PMHS experiments by Kajzer et al. (1997, 1999) (Figures 6 and 7). However, in shearing-type tests at an impact speed of 5.56 m/s, the peak impact forces of both these impactors were higher than those measured on PMHS (Figure 6a). This phenomenon may be related to the following two factors. The first is that the eminence of the meniscus of the new JARI impactor is thicker and broader than that of the human knee joint, which may result in too high stiffness of the knee joint in this impactor. The second factor that could lead to relatively high impact force of the new JARI impactor in shearing-type tests at low impact speed is that the leg of this impactor consists of a virtually rigid tube that directly interacts with the simulated ground, which cannot represent effects of inversion/eversion in the ankle joint on the responses of human lower extremity.

Furthermore, peaks of the impact force-time histories of both the TRL and new JARI legform impactors were delayed in relation to those of the PMHS legs (Figure 6 and 7). This delay was clearly shorter at an impact speed of 11.11 m/s than 5.56 m/s, and similar for both impactors. Therefore, we suggest that the present delay in the peak impact force of the TRL and new JARI impactors could be caused by the low stiffness of the memory foam used as leg padding in these impactors, which could result in a low rate of increase in the impact force. However, the present study alone is not sufficient to verify the validity of this suggestion.

A comparison of the responses between the TRL and new JARI impactors indicated only minor differences between their impact force-time histories (Figures 6 and 7).

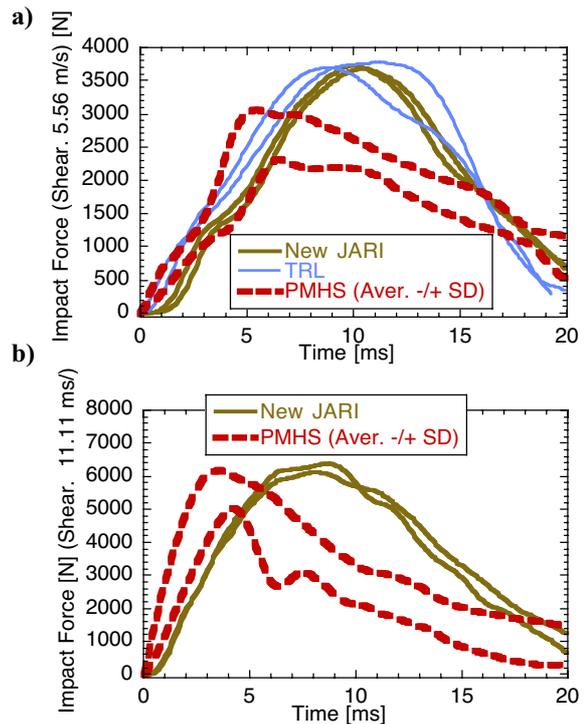


Figure 6. Comparison of impact force-time histories of the TRL and new JARI impactors with PMHS responses. Shearing-type tests. Impact speeds of a) 5.56 m/s and b) 11.11 m/s.

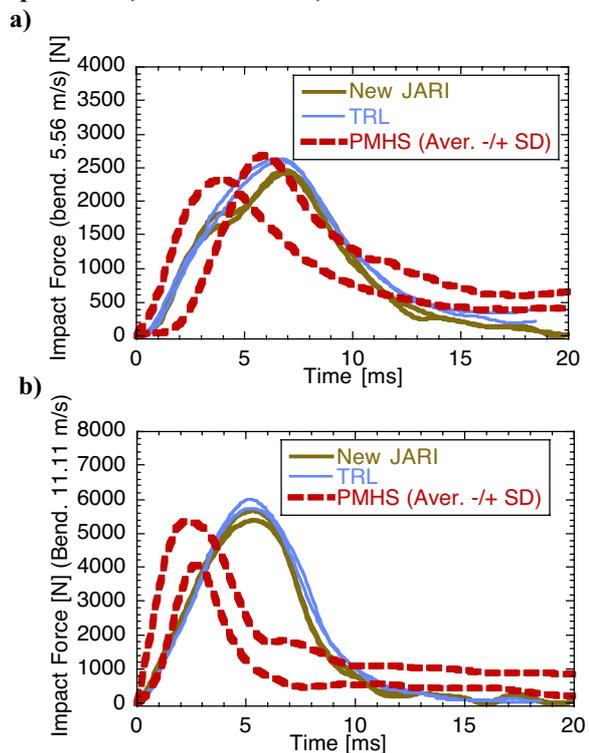


Figure 7. Comparison of impact force-time histories of the TRL and new JARI impactors with PMHS responses. Bending-type tests. Impact speeds of a) 5.56 m/s and b) 11.11 m/s.

Leg Shearing Displacement

In the initial phase of the impact at a speed of 5.56 m/s, the shearing displacement-time histories of the new JARI impactor were within the response corridors determined on PMHS. However, the rate of their increase was lower than that observed on PMHS. In consequence, at an impact speed of 5.56 m/s, the peak value of the shearing displacement of the new JARI impactor leg was around 14 mm, which is clearly below the lower limit of the PMHS responses (Figure 8a). One possible explanation for this phenomenon can be the high stiffness of the knee joint of this impactor

at low impact speed as discussed on page 5. On the other hand, at a speed of 11.11 m/s, the shearing displacement-time histories of the new JARI impactor were very close to the lower limit of the PMHS responses (Figure 8b).

In contrast to the new JARI impactor, the shearing displacement-time histories of the TRL one exhibited a limit of 7.5 mm. Therefore, the shearing displacement-time histories of the new JARI impactor compared to the TRL one were appreciably closer to the PMHS responses (Figure 8).

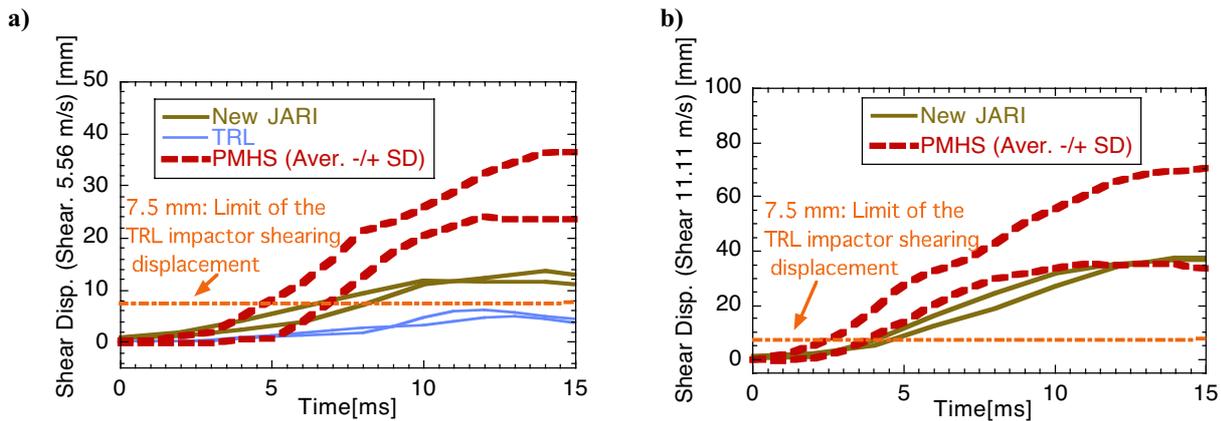


Figure 8. Comparison of shearing displacement-time histories of the TRL and new JARI impactors with PMHS responses. Shearing-type tests. Impact speeds of a) 5.56 m/s and b) 11.11 m/s.

Leg Bending Angle

Shearing-Type Tests In these tests, the bending angle-time histories of the new JARI impactor were very close to the upper limit of the PMHS responses at both impact speeds of 5.56 and 11.11 m/s (Figure 10). However, in case of the PMHS, these time histories exhibited negative values of up to -10° in the initial impact phase, i.e., displacement of the leg proximal part in a lateral direction was greater than that of the distant part, whereas the new JARI impactor yielded a minimum value of the bending angle of only around -2° . One possible explanation for the differences in the bending angle-time histories between the new JARI legform impactor and the human lower extremity may be the following. The impactor leg is virtually rigid, and negative values in its bending angle-time histories result from its rigid-body motion alone (Figure 9b). On the other hand, an appreciable bending-type deformation of the PMHS legs was observed in the experiments by Kajzer et al. (1997, 1999). Thus, in their experiments, negative bending angle could result not only from motion of the leg as a rigid-body, but

also from the deformation of tibia and fibula (Figure 9a).

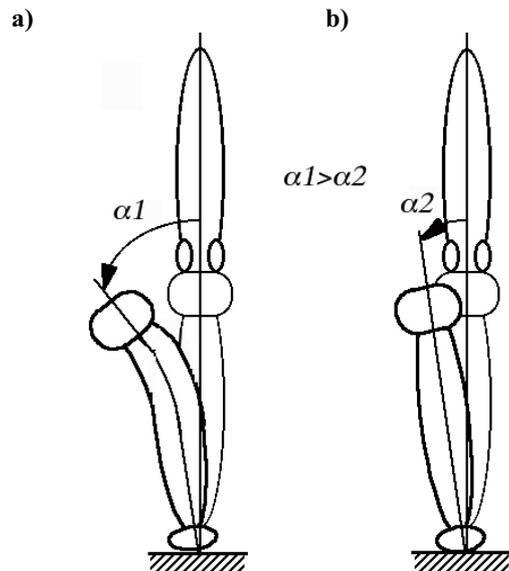


Figure 9. Negative bending angle of a) Human leg (α_1); b) Legform impactor with a rigid leg (α_2).

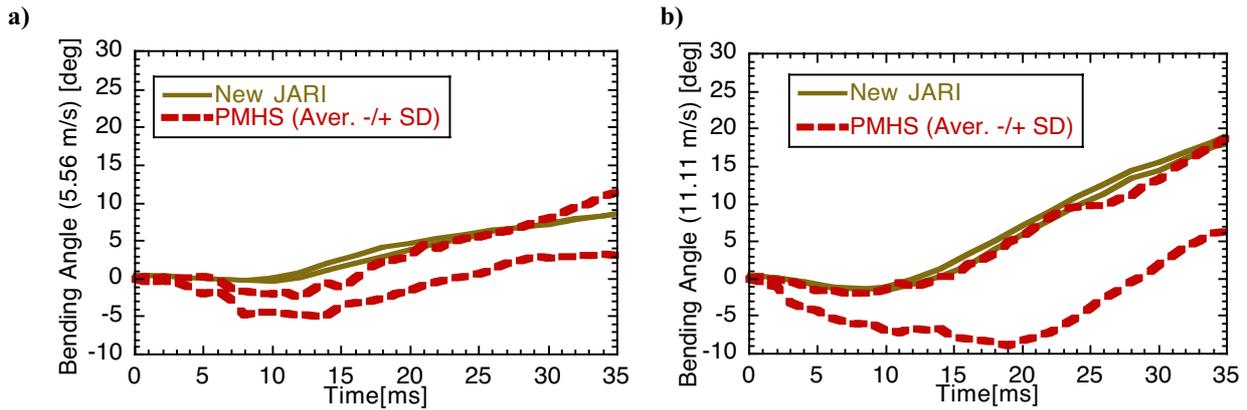


Figure 10. Comparison of bending angle-time histories of the new JARI impactor with responses of PMHS. Shearing-type tests. Impact speeds of a) 5.56 m/s and b) 11.11 m/s.

Bending-Type Tests In these tests, the bending angle-time histories of the new JARI legform impactor were within the PMHS response corridors during the initial 20 and 25 ms of impacts at speeds of 11.11 m/s and 5.56 m/s, respectively. For time values exceeding these initial time-windows, the bending angle-time histories of this impactor were slightly above the upper limits of the PMHS responses (Figure 11).

On the other hand, magnitudes of the bending angle-time histories of the TRL impactor at a speed of 5.56 m/s were lower than those obtained in the PMHS experiments. For a speed of 11.11 m/s, the peak bending angle of the TRL impactor was above the lower limit of the PMHS responses (Figure 11).

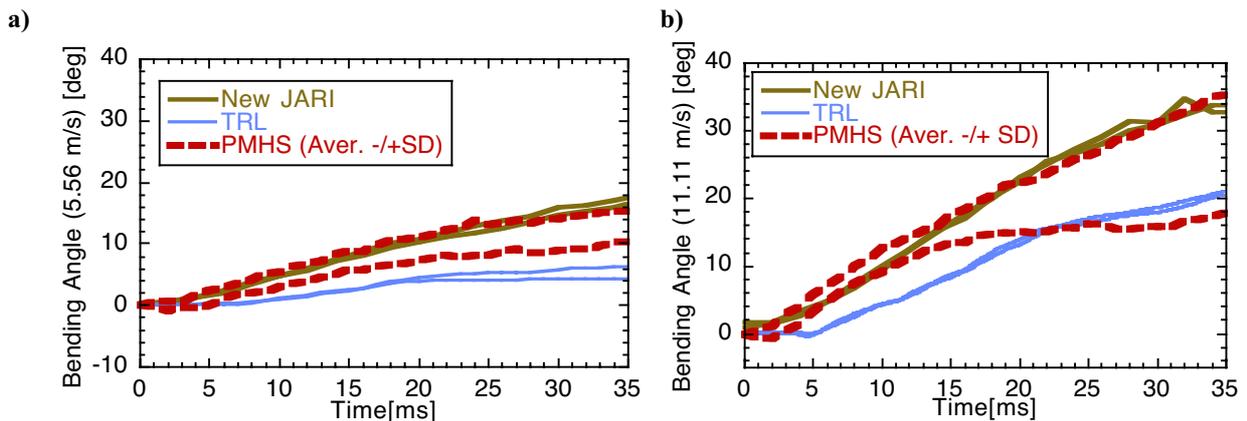


Figure 11. Comparison of bending angle-time histories of the TRL and new JARI impactor with responses of PMHS. Bending-type tests. Impact speeds of a) 5.56 m/s and b) 11.11 m/s.

DISCUSSION

Evaluation of Biofidelity of New JARI Legform Impactor

Responses of the new legform impactor developed in the present study were reasonably close to those of the human lower extremity determined by Wittek et al. (2000) using the results of PMHS experiments by Kajzer et al. (1997, 1999). The peak values of the impact force-time histories of this impactor were within the corridors of the PMHS responses for all the

analyzed tests, except for the shearing-type experiments at a speed of 5.56 m/s (Figures 6 and 7). In these experiments, the peak impact force of the new JARI impactor was around 25% higher than that measured on PMHS (Figure 6a), which might be related to the oversized tibial eminence and disregarding the ankle joint in this impactor.

Furthermore, the peaks of the impact force-time histories of the new JARI impactor were delayed in relation to those determined using PMHS. We suggest that this delay might be caused by the low stiffness of the memory foam we used as padding for the impactor

leg. However, the present study is insufficient to confirm this suggestion. Therefore, we cannot exclude the possibility that the present discrepancies between the impact force-time histories obtained using PMHS and the new JARI legform impactor were also related to factors other than the properties of the leg padding foam. One of such factors could be the characteristics of the contact between the PMHS's foot and simulated ground in the experiments by Kajzer et al. (1997, 1999), which differed from those of the contact between the distal end of our impactor leg and the steel/teflon plates used to represent the ground in the present study.

As with the impact force, the peak values of the shearing displacement of the new JARI impactor were within the corridor of the PMHS responses at an impact speed of 11.11 m/s (Figure 8b). However, at 5.56 m/s, they were below the lower limit of these responses (Figure 8a). One possible reason for the too low shearing displacement of the new JARI impactor in the shearing tests at 5.56 m/s may be differences between the size of the tibial meniscus in this impactor and in the human knee joint as already mentioned in the *DISCUSSION*.

In the bending-type impacts, the bending angle-time histories of the new JARI impactor were within the PMHS response corridors during the initial 20 and 25 ms of impacts at speeds of 11.11 and 5.56 m/s, respectively (Figure 11). In the impact phase following these initial time-windows, the magnitudes of these time histories were slightly above the upper limits of the PMHS responses. This discrepancy between the bending angle-time histories of the new JARI impactor and the human leg seems to be too small to compromise the biofidelity of this impactor. However, it may be of importance when applying the new JARI impactor as a test device since, even at an impact speed of 5.56 m/s (20 km/h), the peak values of its bending angle-time histories exceeded a limit of 15° proposed in the EEVC (1998) report as the acceptance level for legform impactor to bumper test.

In shearing-type impacts, the bending angle-time histories of the new JARI impactor were also very close to the upper limit of the PMHS responses (Figure 10). However, in the initial phase of shearing-type impacts, considerably negative values of the bending angle were observed in the motion of the PMHS's legs, whereas for the new JARI impactor, the minimum bending angle was only around -2°. The likely reason for this phenomenon is that the leg of this impactor is virtually rigid, whereas a notable bending-type deformation of the human leg was observed in the PMHS experiments by Kajzer et al. (1997, 1999). Our reason for disregarding the leg deformation was that the application of a rigid leg is simple and commonly

accepted in design of legform impactors, e.g., the TRL impactor (EEVC, 1998).

Comparison of Biofidelity of TRL and New JARI Legform Impactors

Despite only minor differences in the impact force-time histories of the TRL and new JARI impactors, their kinematics differed significantly. Magnitudes of the shearing displacement and bending angle-time histories of the new JARI impactor were appreciably higher and closer to the PMHS responses than those of the TRL impactor. Thus, it can be concluded that the design of the complex structure simulating the geometry and mechanical properties of the human knee joint in the new JARI impactor made it possible to achieve biofidelity exceeding that of the other legform impactors reported in the literature.

Recommendations for Further Studies

The present investigation indicated that, although the peak values of the impact force-time histories of the TRL and new JARI legform impactors are close to those determined in the experiments on PMHS, they are delayed in comparison to the PMHS histories. This delay is very similar for both these impactors despite differences in the structure and geometry of their knee joints. This, in turn, suggests that the delay is not related to the features of their knee joints as such, but rather to the properties of contact between the side ram and legs of these impactors. As discussed earlier (see page 5), these properties can be determined by the low stiffness of the memory foam (Confor™) utilized as leg padding in both the TRL and new JARI impactors. We did not attempt to determine to what extent the properties of this foam are similar to those of human flesh since its application was recommended in the proposed European standard (EEVC, 1998). However, our experience in conducting experiments using the Confor™ foam suggests that its stiffness is very sensitive to temperature and may exhibit some strain-rate dependency. Therefore, we recommend that a parametric study be done to determine if an increase in the static stiffness of leg padding reduces the delay in the impact force-time histories of the new JARI and TRL legform impactors.

As already suggested in the *RESULTS* and *DISCUSSION*, one possible reason for the differences in the bending angle-time histories between the new JARI impactor and the PMHS lower extremities under a shearing-type load may be the use of a virtually rigid tube to represent a leg in this impactor. A direct way to confirm validity of this suggestion is to add a deformable leg to the new JARI impactor.

Furthermore, in bending-type impacts, the leg bending angle-time histories of the new JARI impactor were slightly above the upper limits of the PMHS responses for time values exceeding 20 and 25 ms at impact speeds of 11.11 and 5.56 m/s, respectively. One possible solution to reduce the magnitude of these time histories can be to increase the stiffness of springs of the knee joint ligaments.

Our recommendations for further studies are based on hypotheses regarding the causes of the present differences between the responses of the new JARI legform impactor and a human lower extremity. However, the present study was designed to evaluate the general biofidelity of this impactor, and it does not enable us to thoroughly confirm these hypotheses. The validation of our suggestions for improvement of biofidelity of the new JARI legform impactor requires a new experimental investigation.

CONCLUSIONS

In the present study we developed a new legform impactor (referred to as the new JARI impactor) and evaluated its biofidelity. The responses of this impactor were close to those of the human leg. In bending-type impacts at speeds of 5.56 and 11.11 m/s and in shearing-type impacts at a speed of 11.11 m/s, the peak values of its shearing displacement and its bending angle-time histories were nearly within the corridors determined in the experiments using PMHS. These results imply that the new JARI legform impactor rather than the TRL impactor more accurately represents the kinematics of the human leg in lateral impacts.

It is important to determine the cause and reduce the differences between the responses of the new JARI impactor and human lower extremities. One such difference is delay in the peak of the impact force of this impactor and too low magnitude of negative values in its bending angle-time histories in the initial phase of shearing-type impacts. To achieve this we suggest the following: 1) To conduct lateral impact experiments using an impactor leg padding with a higher static stiffness than the currently used Confor™ foam; and 2) To replace the rigid impactor leg with a deformable one.

ACKNOWLEDGEMENTS

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