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A Biomechanical Method for Reconstruction of Tumbling Trampoline-Associated Cervical Spine Injuries Using Human and Anthropometric Test Dummy Data

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Abstract

Background

Rebound devices such as trampolines are associated with catastrophic spinal cord injuries. Cadaveric studies have reported thresholds for injuries that can be applied to the case of failed acrobatics such as backward somersaults. However, it remains unclear whether falls on rebound surfaces should be expected to cause neurological injuries in the majority of cases or only in unfortunate exceptions. The purpose of the current study is to demonstrate the risk of injury associated with a failed backflip performed on a rebound device such as a trampoline or tumbling trampoline.

Methods

Backward somersault kinematic data was acquired using subjects fitted with a safety harness. This data was then applied to a testing rig designed to set a Hybrid III Anthropometric Test Dummy (ATD) in rotation and released at precise timing so as to reproduce failed backward somersaults. The ATD was instrumented to measure rotational rate, head acceleration as well as stresses in the lower cervical region.

Results

The axial compression, shear force and flexion moment measured on average (SD) were respectively 1700 (470) N, 909 (667) N, and 360 (122) Nm while the threshold for bilateral facet joint dislocation (BFD) demonstrated by previous cadaver studies showed a significantly lower threshold ($p<0.001$). Combined results have shown a likelihood of BFD for failed somersaults on tumbling trampolines ranging from 47 to 99%.

Conclusion

Failed backward somersaults causing BFD are also likely to cause neurological damage. Therefore, use of rebound devices requires the need for progressive skill achievement; supervision for beginners and the use of additional safety measures must be emphasized to prevent inverted vertical falls resulting in the specific combination of forces necessary to cause BFD.

Introduction

Every year, thousands of people report to hospital emergency departments for trampoline related injuries [1-3]. Out of all these injuries, 8 to 12% are spinal injuries and of those injuries, approximately 5% result in permanent damage to the patient's neurological functions [2,4].

Although many trampoline accidents involve accidental contact with the trampoline frame or excursions outside of the rebound surface, many of the spinal injuries which result in neurological damage occur within the parameters of regular trampoline use [5]. Multiple case studies of failed backward somersaults show the inadequacy of trampolines to protect the lower cervical spine from catastrophic injuries [6].

In order to understand the likelihood of such injuries and demonstrate the risk involved in the unsupervised use of such devices, a biomechanical study was carried out using a two-step approach. To accurately reproduce the conditions of such incidents, the kinematics of backward somersaults was acquired using beginner, intermediate, and expert gymnasts. The kinematic data was then applied to an

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Anthropometric Test Dummy (ATD) instrumented to measure cervical loads.

Background

The risk associated with trampolines and other rebound devices has long been known [6]. Reports of fractures and fracture-dislocations have become quite common [7]. However, serious neurological damage can occur at much lower stresses when dealing with Bilateral Facet Dislocations (BFD) [3,8,9]. The risk of dislocations can therefore be considered as a minimal threshold for possible catastrophic spinal injuries when looking at trampolines. The potential for spinal cord lesions and quadriplegia in the case of BFD has been reported to be between 50 to 84% [10,11].

Ivancic et al. [12] have shown such injuries to occur when the spinal cord is subject to a combination of loads as low as 264.5 N in compression, 54.5 N in shear and 17.7 Nm in flexion for the C5/C6 spinal unit. The threshold of neighbouring spinal units was reported as being slightly higher in flexion moment as shown in Table 1, while other stress types were comparable.

Loading	C3/C4	C5/C6	C7/T1
Axial Compression (N)	215.5	264.5	226.7
Anterior Shear (N)	60.6	54.5	108.3
Flexion Moment (Nm)	35.3	17.7	37.4

Table 1: Neck load tolerance values as reported by Ivancic et al. [12].

Nightingale et al. also observed this type of injury in cadaver head and neck specimen that had been rotationally constrained under axial loading [8]. It may be observed that this type of constraint is comparable to the behaviour of a head being pushed into a compliant surface. The values obtained by this research group, however, differ from the Ivancic findings [12] by showing a more conservative injury threshold of 1720 ± 1234 N (mean ± SD) for axial loads exclusively. A similar study by the same research group has also shown two dislocations caused by pure flexion moments with thresholds of 36.2 Nm at C1-C2 and 42.2 Nm at C2-C3 [13]. Although these results cannot be directly compared to Ivancic's results on a spinal unit basis, the order of magnitude of these results are similar to the most caudal spinal units reported by Ivancic [12].

The results published by the two research groups offer a range in which the BFD can be expected to occur. This can therefore be applied as injury threshold when interpreting the results of an analogous model.

Further, this type of injury occurs quickly according to Winkelstein and Myers' 1997 review paper [14]. By imaging the mechanism as it took place, cervical spine injuries have been shown to occur in less than 20 ms.

Materials and Methods

Human surrogate testing

Backward somersault kinematic data was acquired by using 8 test subjects of varying skill level, height and weight whose informed consent was obtained. Table 2 summarizes the biometrics and skill level of each subject. The former was assessed by an expert in gymnastics [15].

Subject ID*	Weight (kg)	Height (cm)	Gender (M/F)
B1	59.5	170.8	M
B2	69.7	175.3	M
B3	73.3	176.3	M
B4	64.0	185.9	M
B5	54.9	171.5	M
I1	77.7	182.6	M
E1	65.5	170.7	F
E2	63.4	165.1	F
E3	53.8	159.0	F

Table 2: Subject biometrics and gymnastics proficiency. *B: Beginner; I: Intermediate; E: Expert.

Subjects were instrumented using light-reflective markers as to track rotation of their torso and head (Figure 1).

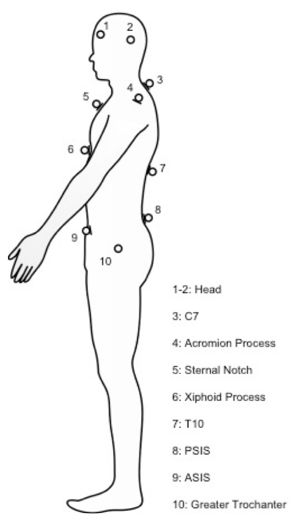

Figure 1: Marker placement (PSIS: Posterior Superior Iliac Spine and ASIS: Anterior Superior Iliac Spine).

For improved safety, the subjects were placed into a gymnastic harness, which was attached to an overhead traveling rig installed above the rebound device. The harness allowed freedom of movement to jump and rotate while allowing a spotter to prevent accidental contact of the head or neck on the rebound surface.

Each subject was asked to perform at least 3 backward somersaults under each of the following conditions: using 2 preliminary bounces, using 1 preliminary bounce, using no preliminary bounce. By default, beginner rotation rates were not fast enough to have successfully completed the somersault without the aid of a harness. All somersaults were recorded using a high-speed Photron Camera positioned laterally as to observe motion in the sagittal plane. The footage was then processed to extract the head and limb angles using the “Normal Angle” tool of Silicon Coach Software. The “Stop Watch” function was also used to identify the time at which the subject reached the “inverted position”. This time was used as the “time of impact”.

Head and limb angle were measured using the reflective markers at a sampling rate of 100 Hz as shown in Figure 2. Angular rate of change was measured and filtered using a 9-point moving average.

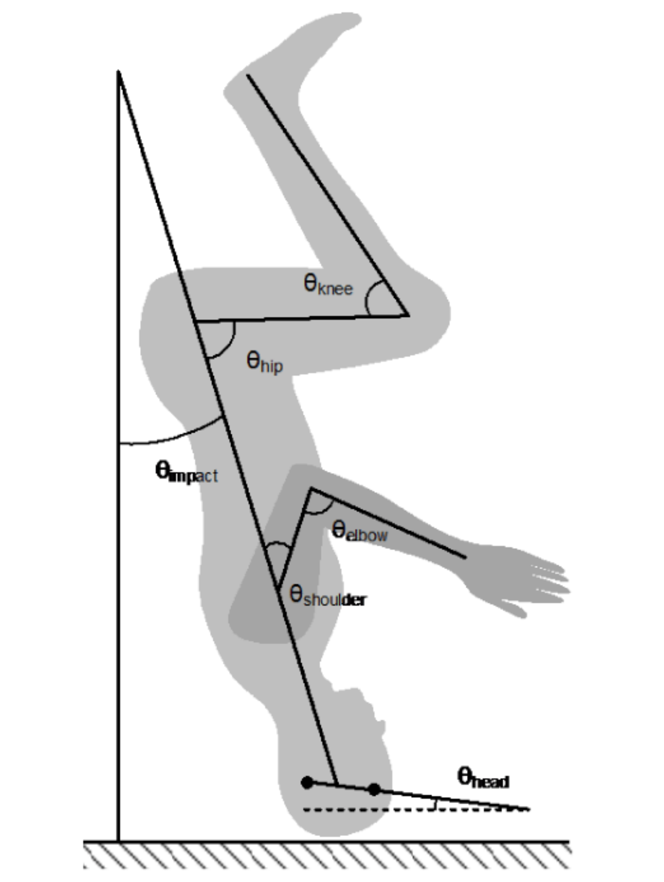

Figure 2: Angle measurement convention.

To measure jump height, one subject was instructed to perform vertical jumps in the same manner they would if they intended to perform a backward somersault. Jumps were performed using each of preliminary jump conditions previously described. The Stop Watch function was then used to measure the subject's airtime, from toe off to toe on, and calculate the jump height based on the equation of motion of a free falling body strictly under the effects of gravity.

The rebound device used for this test was a tumbling trampoline (TT) manufactured by TumblTrak (Mount Pleasant, MI), 30 ft in length and 6.9 feet in width.

ATD Drop testing

Using the kinematic data obtained through the less experienced human surrogate testing, the conditions of a failed backward somersault injury were reproduced using a fifty percentile male Hybrid III anthropomorphic test dummy (Humanetics ATD, 78051-218-H, Huron, OH). To conservatively reconstruct falls suffered by the less experienced, the lower beginner rotation rates were used together with only the lower jump heights recorded (no preliminary bounce condition).

To effectively reproduce the kinematics of a full-scale failed backward somersault, a testing rig was designed and constructed in order to achieve repeatable, predictable testing results (Figure 3).

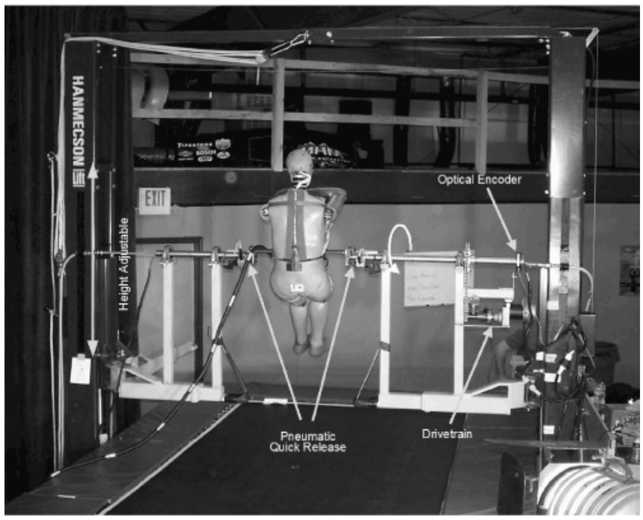

Figure 3: ATD testing rig.

This rig used an electric motor coupled to an optical encoder, with a closed-loop feedback control system. The encoder was mounted directly to the axle that held the dummy in order to achieve a desired rotational speed of the system. The ATD was mounted on a pneumatic quick-release axle coupled with the geared motor. A hydraulic lift was used to control the height of the axle from which the rotating dummy would drop. The system was controlled via software where the rotation speed could be programmed and monitored relative to the pneumatic quick-release of the dummy. During each test, the dummy was rotated to a prescribed angular rate and then released at a precise moment such that the dummy would rotate, fall, and contact the trampoline surface at a desired orientation.

An angular rate sensor (Diversified Technical Systems, ARS-1500, Seal Beach, CA) positioned in the posterior portion of the pelvic region was also used to measure the torso rate of rotation. A standard CFC180 filter was then applied to the signal [11]. The same laterally positioned camera setup was used to measure the head angle at impact and confirm the rotational rate measured by the angular rate sensor. Markers were positioned on the ATD, in a similar manner as with the human subjects as to extract head angle at impact and rotational rate.

To describe the loads applied to the cervical spine, a 6-axis load cell (Humanetics ATD, IF-210-HC, Huron, and OH) was positioned at the junction between the neck and thorax of the dummy. This junction can be assumed to represent the approximate location of the C5/C6 spinal unit. Acceleration of the head was measured by two sets of 3 linear accelerometers (Endevco, #7264C-2K, Irvine, CA) positioned orthogonally at the head's center of gravity. Head Injury Criterion or HIC (HIC15 and HIC36) was then calculated for each dataset. All data collected by the load cell and accelerometers was filtered using a standard CFC1000 filter, in accordance with SAE J211.

Tests with the ATD were carried out on the same tumbling trampoline as for human testing. For comparison purposes, the same ATD tests were repeated on a high performance trampoline (6 ft by 12 ft). This trampoline was qualitatively observed as being more compliant than the tumbling trampoline.

Groups differed between skill levels and surfaces. Descriptive statistics and independent samples t-tests between groups were applied to the data. Risk of injury was quantified using a cumulative probability plot based on the comparison of study results and previously published injury thresholds. Linear regression was used to determine relationships between biomechanical injury parameters and somersault kinematics.

Results

Human surrogate testing

The average jump heights were measured at 0.75, 1.03, and 1.09 ± 0.01 m for standing, one, and two bounces respectively. Limb orientations were also measured for each jump and averaged as shown in Table 3. The average was subsequently used for the ATD drop test.

Joint	Avg Angle ± SD (deg)	Range (deg)
Elbow Angle Θ1	115 ± 10	45 to 180
Shoulder Angle Θ2	19 ± 13	-41 to 178
Hip Angle Θ3	65 ± 4	21 to 83
Knee Angle Θ4	76 ± 7	39 to 127

Table 3: Joint angles at impact, measured in the sagittal plane.

Rotational rates varied greatly amongst skill levels. However, expert gymnasts (E1, E2) showed higher rotational speeds than beginners and intermediate gymnasts with an average of 515 ± 55 deg/sec and a range of 354 to 667 deg/sec as shown in Figure 4.

In comparison, the results of the more novice subjects (B1-B5, I1) averaged at 291 ± 14 deg/sec with a range of 149 to 455 deg/sec. An independent samples t-test performed on these results shows expert rotational rates as significantly higher compared to the less experienced gymnasts and, therefore, should not be used interchangeably (p<0.0001). The rotational rate results of the less experienced were therefore chosen as the range to use for ATD trials.

ATD Drop testing

A total of 44 trials were performed. The body angle at impact for these simulated failed backward somersaults averaged at 13.5 ± 17.4 deg with a range of -29 to 47 deg with respect to the vertical.

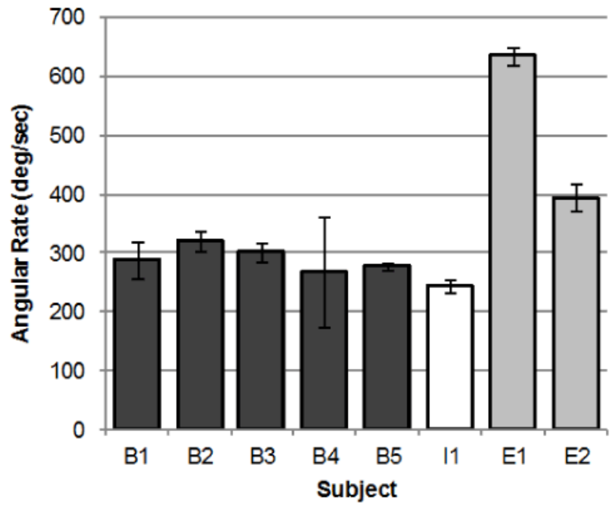

Figure 4: Average angular rate at impact grouped by skill level. Under an independent samples t-test, the average of the expert (E1, E2) data was shown to be significantly greater (p<0.0001) when compared to the rest of the data.

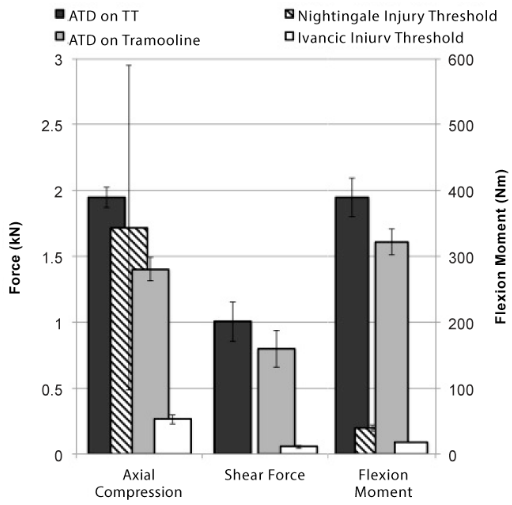

Figure 5: ATD maximal loads for each axis.

The measured loads are displayed in Figure 5 along with the C5/C6 bilateral facet dislocation tolerance values reported by Ivancic et al. [12] and Nightingale [8,13].

The peak values collected during the current study can be seen breaching the Ivancic threshold on average 24.0 ± 5.1 msec and 23.0 ± 4.8 msec after impact for the TT and high performance trampoline data, respectively.

Figure 6 shows neck load data in relation with rotational rate. The data is largely concentrated between 200-300 deg/sec. The relationship is non-linear but indicates higher loads for lower angular rates.

Discussion

The purpose of this study was to investigate the likelihood of cervical injury for failed backward somersaults on rebound devices such as trampolines and tumbling trampolines by using BFD as a benchmark.

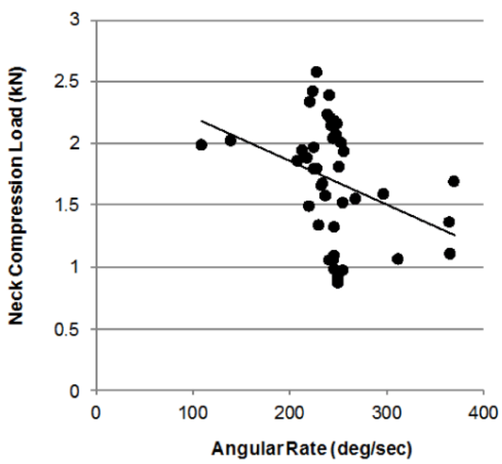

Figure 6: Neck compression load vs. angular rate at impact point.

The HIC values recorded for the entire study ranged from 3 to 32. These values are well below the allowable limit of mid-sized males for automobile accidents which stands at 1000 [16].

The results collected on the TT show compression loads on average 7 fold higher (from 3.3 to 9.7) than the injury threshold of a C5/C6 spinal unit reported by Ivancic [12]. The results observed on the trampoline also breach the injury threshold but at lower values, which is expected considering the higher compliance of the high performance trampoline over the TT. One may also wish to compare the values obtained with the Nightingale injury threshold of 1720 N for a rotationally constrained head [8,13]. In this instance, the results are more closely matched but still significantly different between the two devices with average stresses of 1949 ± 79 N (TT) and 1401 ± 90 N (trampoline).

Statistically speaking, the likelihood of injury can be quantified using a cumulative probability plot based on the comparison of study results and previously published injury thresholds. In the case of the Ivancic thresholds, the results show a 99.99% probability of occurrence for breaching the threshold for all three measured stresses. Meanwhile, the Nightingale threshold shows a likelihood of injury of 47.01% based on its axial compression threshold. In other words, one may consider the likelihood of injury for a failed backward somersault on a rebound device to stand between 47.01% and 99.99%.

According to the results shown by Ivancic, the threshold loads to cause disarticulation of the spine at C5-C6 are accompanied with a narrowing of the spinal canal. This canal pinch diameter has been measured post injury to be 2.2 ± 0.8 mm in terms of average and standard deviation. This relatively small compression does not explain the spinal cord lesion that typically occurs [17]. What does explain the spinal cord lesion is the follow through loading or dynamic loading that continues to increase the canal pinch diameter roughly three times that of post impact narrowing [17] at 6.4 ± 3.6 mm. Figure 7 illustrates the increased pinch suffered by the spinal cord through the three key phases of the injury.

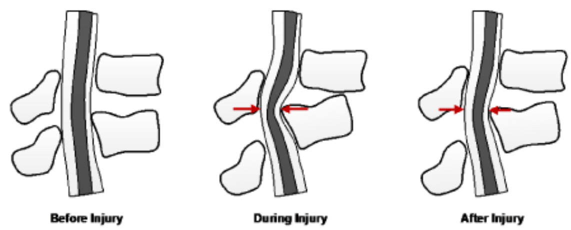

Figure 7: BFD damage to the spinal cord.

Moreover, the time required for all three of Ivancic's injury thresholds to be exceeded stands at 24 and 23 msec on average for the TT and trampoline respectively. This is in close agreement with the cited articles [14,18] since the current experiments were performed on a more compliant surface that would tend to increase the time to peak load.

Further, cervical spinal cord injuries often result from the incapacity of the neck to stop the mass of the moving torso. Experiments show that neck injuries can be produced with only a percentage of torso weight, approximately 16 kg, following the head and neck in an impact with a velocity of 3.1 m/s which corresponds to a drop height of about 0.5 m which is lower than the minimal value drop test of 0.66 m used in the current study [9,18].

The type of surface also has an effect on neck injury [19]. In some head impacts, the head and neck are able to bend out of the path of the momentum of the torso that follows; the torso (chest, shoulders, or back) then contacts the surface and absorbs the torso momentum without loading the neck. This sort of behaviour is made less likely when dealing with rebounding devices such as a TT or a trampoline because of the reaction of the surface to the contact. As previously explained by Winklestein and Myers the pocketing of the head by a compliant impact surface reduces the ability of the head and neck to move out of the way of the torso and can increase the risk for neck injury [14]. This behaviour of the contact surface can be observed in the high speed videos of the simulated ATD falls as shown in Figure 8 which illustrate this point on a TT.

Additionally, the relationship seen between neck load and angular rate further confirms that a beginner performing a backward somersault at a lower angular rate would be more susceptible to suffer from the pocketing effect. To understand this, one may consider the opposite case where, at higher rotational velocities, the additional momentum of the body would tend to reduce the duration at which the inertia of the torso is aligned with the spine thus limiting the pocketing effect.

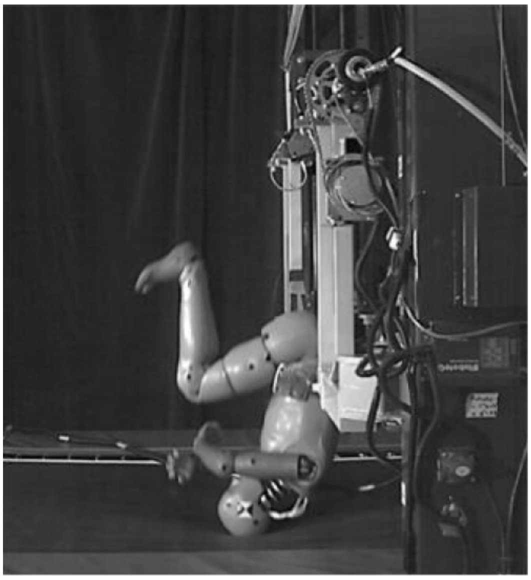

Figure 8: Head pocketing as reproduced during ATD testing.

In summary, five published mechanisms of injury agree with the test data: 1) Neck loads were sufficient to exceed reported thresholds to cause bilateral facet joint dislocation of C3/C4, C5/C6 or C7/T1 which caused significant canal pinch diameter; 2) Time to reach these threshold levels correspond well with the literature for the given conditions; 3) Minimal drop heights determined by human surrogate tests exceed reported drop heights that can cause cervical neck injury; 4) With this lower drop height, only a fraction of body weight was required to cause neck injury; 5) "Pocketing" of the impact surface such as that of rebound devices such as trampolines and TT increases risk of neck injury.

Conclusion

With a likelihood of injury spanning from 47 to 99%, vertical falls on trampolines and tumbling trampolines where the head is the primary point of contact represent an undeniable risk. The need for progressive skill achievement, supervision for the less experienced and the use of additional safety measures must be emphasized. Such measures are readily available in the form of spotting rigs and harnesses such as the ones used to acquire human kinematic data in the current study. With the help of a qualified coach and spotter the risk of head contact caused by a failure to complete a somersault manoeuver is practically eliminated, thus eliminating the potential for injury.

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Nomenclature

ATD: Anthropometric Testing Dummy

BFD: Bilateral Facet Dislocation.

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