The effect of acceleration signal processing for head impact numeric simulations

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Abstract

Brain injury research in sport employs a variety of physical models equipped with accelerometers. These acceleration signals are commonly processed using filters. The purpose of this research was to determine the effect of applying filters with different cutoff frequencies to the acceleration signals used as input for finite element modelling of the brain. Signals were generated from reconstructions of concussion events from American football and ice hockey in the laboratory using a Hybrid III headform. The resulting acceleration signals were used as input for the University College Dublin Brain Trauma Model after being processed with filters. The results indicated that using a filter with a cutoff of 300 Hz or higher had little effect on the resulting strain measures. In some cases there was some effect of the filters on the peak linear (8–30g) and rotational measures (1000–4000 rad/s2), but little effect on the finite element strain result (approximately 2–6 %). The short duration and high magnitude accelerations, such as the puck impact, were most affected by the cutoff frequency of different filters.

Keywords Signal processing, Filtering, Concussion, mTBI, Biomechanics, Ice hockey, American football

1 Introduction

Brain injury research often involves linear and rotational acceleration signals to quantify the risk of incurring an injury, be it a traumatic brain injury (TBI) or concussion (mTBI) [1–4]. Typically, to acquire these acceleration signals accelerometers are used that are installed on either a helmet for real-time acquisition of game data [5, 6], or inside of anthropometric dummy systems such as the Hybrid III [4, 7, 8] or NOCSAE headforms [9]. This data collection for brain injury is based upon car crash conventions, most often the SAE J211. This convention specifies the rate of data collection (minimum 10 kHz) and the filter used for head impact research using an anthropometric dummy, a CFC (Channel Frequency Class) filter. This convention is applied to ensure that the sampling rate is high enough to gather the information from an impact. The filtering convention is applied because often signals from accelerometers can be noisy and have sensor drift; this type of signal processing can minimize those effects on the resulting data. Filters can have a large effect on the signal and as a consequence can give misleading results if the effect of this type of signal processing on the raw signal is not well understood, particularly for sporting impact research. However, while some research in the literature follow the SAE J211 convention [2–4, 10], several use more aggressive filters [11, 12] and in some cases signals are filtered twice [5]. In addition to these inconsistencies in data treatment, the SAE J211 convention was developed for car crash analyses and not for processing data for sporting impacts. The effect signal processing might have on results is currently unknown and could affect comparisons between research using dissimilar data treatment techniques.

Common brain injury acceleration data collection in the literature follows one of two avenues, each with its own unique limitations and considerations. Those two methods are in-helmet sensor datasets and the other is from Hybrid III headforms with accelerometers installed inside. The in-helmet sensor data are commonly collected at 1000 Hz, and in many of the publications that use these types of systems a description of the data filtering method is not reported [5, 6]. However, in other publications using the same type of head impact systems, it was reported to be filtered using a 400 Hz low-pass filter [12]. When examining the research that uses Hybrid III headforms, there is some variation in the application of the commonly used low-pass filter. Beckwith et al. [5] report the use of a low-pass filter with a 1650 Hz cutoff for the linear accelerations, and a more aggressive 300 Hz low-pass filter on the rotational accelerations. Other researchers have published extensively using accelerometers installed in a Hybrid III headform [2–4, 13] both for brain injury research and helmet testing and report using a lowpass filter with a 1650 Hz cutoff, which is a higher cutoff frequency than the SAE J211 convention.

The only literature that compares the effects of filtering on brain injury results was published by Newman et al. [14], who analyzed the data collection for a series of NFL American football impacts that resulted in concussion. Newman et al. investigated the effect of filtering accelerometer data using low-pass filters with 1650 and 300 Hz cutoffs, resulting in an 8.1 % decrease in peak linear and rotational acceleration responses [14]. While this research showed that filtering had an effect on the results of brain injury reconstructions, the most appropriate filtering frequency as determined through a Fourier analysis was not discussed. Recent research has employed extensive use of finite element models of the human brain to determine the brain stresses and strains from impact, as these metrics have been found to be more closely linked to concussion [15–17]. The results from these brain injury FE models are directly related to the shape and characteristics of the linear and rotational acceleration loading curves that are the product of impacts to instrumented Hybrid III headforms that underwent some sort of filtering process [18, 19]. This research makes use of filter specifications not designed for this type of head impact research, but rather for signals generated from car crash simulations. As a result, using a filter may have unintended effects on the signals produced from head impact reconstruction. The use of different filters may affect the output of the linear and rotational acceleration time histories and make comparisons between research using different signal processing techniques difficult. Since the FE results are dependent on curve shape (signal magnitude, duration, slope) [19], and the filters affect these characteristics, it is important to know how this signal processing affects the results of brain injury investigations using these tools and may help inform standard committees when developing testing methodologies. The purpose of this research is to examine the effect of low-pass filters on the linear and rotational acceleration signals generated from impacts to a Hybrid III headform in an effort to determine how this signal processing can affect measures of strain as measured through a finite element model of the human brain.

2 Materials and methods

Nine different concussive events in elite American football and ice hockey were reconstructed to obtain signals that would be representative of a Hybrid III brain injury research methodology. Of these nine, four were common American football events of injury: knee to helmet, shoulder to helmet, helmet to helmet, and falling to ground. The remaining five concussion reconstructions were ice hockey events: elbow to head, shoulder to head, fall to ice, head to boards, and puck to head [20]. The cases were identified as concussions from team medical doctor reports from impacts that occurred during competitive play. Once the impacts were identified and confirmed as concussion events, the video for the impact was analyzed using Kinovea 0.8.2 software to determine the characteristics of the impact including velocities, locations, impact vector (angle of contact), mass, and compliance. The specifics of these analyses have been reported in Rousseau [21], Post et al. [22], and Rousseau and Hoshizaki [23].

2.1 Equipment

2.1.1 Hybrid III headform and unbiased neckform

All reconstructions were conducted using a Hybrid III 50th headform (mass 4.54 ± 0.01 kg). It was outfitted with a 3-2-2-2 accelerometer array for the measurement of linear and rotational acceleration [24]. The accelerometers used were Endevco 7264C-2KTZ-2-300 (Irvine, CA, USA) and were sampled at 20 kHz. The signals were collected and processed by computer and DTS TDAS system (DTS, Seal Beach, CA, USA). As the Hybrid III neckform is asymmetrical and can influence the results of a direct head impact, an unbiased neckform was used for the reconstructions. This neckform comprised alternating aluminum and rubber discs. The response of this neckform is symmetrical in all axes and therefore did not bias the results of the impact [25].

2.1.2 Monorail

The falling and head to boards events were reconstructed using a monorail drop rig. The monorail was 4.7 m long and had a drop carriage attached to it by ball bushings. The Hybrid III headformand unbiased neckform were attached to the drop carriage and the drop mechanism was a pneumatically controlled release lever. The impact velocity was determined by laser timegate positioned within 0.02 m of the impact. The anvil was changed depending on the impact surface as determined from video analysis. For the American football falling reconstructions an Modular Elastomer Programmer (MEP) (60 ShoreA)was used to simulate the grass, and for the ice hockey falls ice was frozen at -30 _C and impacted. For the impacts to the boards, a section of ice hockey rink boards was used as the impact surface.

2.1.3 *Linear impactor*

The linear impactor was used to reconstruct the helmet to helmet, shoulder to head, and knee to head impacts. The linear impactor had two parts, the impacting armand the frame. The impacting arm had an air canister, frame, and an aluminium impacting shaft (13.1 kg). At the end of the impacting shaft was an impactor cap that was changed depending on the event that was being reconstructed. The helmet to helmet impacts utilized a Vinyl Nitrile (VN) 602 with a hemispherical nylon cover as the impactor cap. This design of cap has been demonstrated to have similar responses to those of American football helmets [7]. For the shoulder to head impacts, a compliant VN impactor cap was used that was designed to have a similar compliance to that of a human under an impact condition [23]. For ice hockey, the VN cap was used with an ice hockey shoulder pad, and for American football, the VN cap was covered with an American football shoulder pad. For the knee to head reconstructions, a 60 shore A MEP impactor was used to reflect the increased stiffness that would be representative of a knee. The sliding table (mass 12.78 ± 0.01 kg) allowed for movement of the Hybrid III head and neck post-impact by sliding backward following impact. The table was adjusted to allow for the proper positioning of the headform in 6 degrees of freedom.

2.1.4 High velocity impactor

To reconstruct the elbow to head impacts, a high-velocity impactor was used that included a carriage attached to a 3.35 m-long dual rail by ball bushings and projected down the rails. A 3.15 kg anvil with compliance determined through human impact testing [23] was used to replicate a human elbow. The carriage and elbow were released to impact the Hybrid III headform that was attached to the same table as used for the linear impactor.

2.1.5 *Pneumatic puck launcher*

The pneumatic puck launcher consisted of two parts: the frame which held the launcher and air canister and a table that housed the Hybrid III headform and was attached through the unbiased neckform. The target velocity was achieved by regulating the amount of air pressure in the canister. The table was the same as described in the linear impactor description.

2.1.6 *University College Brain Trauma Model*

The University College Brain Trauma Model (UCDBTM) was used to calculate the maximum principal strain (MPS) from the dynamic response values obtained during the reconstructed impacts to the headform [26, 27]. The maximum principal strain was used as the target-dependent variable as a result of its common usage in brain injury research, in particular concussive research [1–4, 16, 28]. The geometry of this model was developed from medical imaging of a male cadaver head and comprised the following parts: scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), gray and white matter, cerebellum, and the brainstem [26, 27]. The UCDBTM was validated against data from Nahum et al's [29] and Hardy et al's [30] cadaveric impact research. In addition, this model has been used for human brain injury reconstructive research and has produced data consistent with that from anatomical testing [4, 28, 31]. In total, the UCDBTM has approximately 26,000 hexahedral elements.

The UCDBTM had material characteristics that were determined from anatomical testing on cadavers and tissue samples [32–36] (Tables 1, 2). The modeling of the response of the brain tissue was conducted by using a linearly viscoelastic model combined with a large deformation theory. The behavior of the tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [26]. The compression of the brain tissue was considered to be elastic. The shear characteristics of the viscoelastic brain were expressed by:

$$(1) \quad G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

With G_∞ representing the long term shear modulus, G_0 the short term modulus and β is the decay factor. A Mooney–Rivlin hyperelastic material model was used for the brain to maintain these properties in conjunction with a viscoelastic material property in ABAQUS, giving the material a decay factor of b = 145 s-1 [26]. The hyperelastic law was given by:

$$(2) \quad C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)}$$

where C_{10} is the mechanical energy absorbed by the material when the first strain invariant changes by a unit step input and C_{01} is the energy absorbed when the second strain invariant changes by a unit step [37, 38] and t is the time in seconds. To define the sliding interface between the brain and skull, the UCDBTM used a sliding boundary condition that defined no separation between the pia and the CSF. The CSF was modeled using solid elements with a bulk modulus of water and a low shear modulus [26, 27] using a coefficient of friction of 0.2 to define the interaction between the two surfaces [39].

Table 1 Material properties of the UCDBTM

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m^3)
Scalp	16.7	0.42	1000
Cortical Bone	15 000	0.22	2000
Trabecular Bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx and Tentorium	31.5	0.45	1130
Brain	Hyper Elastic	0.499981	1060
CSF	15 000	0.5	1000
Facial Bone	500	0.23	2100

Table 2 Material properties of the UCDBTM for the brain tissues

	Shear modulus (kPa) G_0	G_∞	Decay constant (s^{-1})	Bulk modulus (GPa)
Grey matter	10	2	80	2.19
White matter	12.5	2.5	80	2.19
Brain stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

2.2 Procedure

Reconstructions of the concussive events were conducted in the laboratory using the parameters obtained from video. The headform was equipped with either an ice hockey helmet or American football helmet, depending on the sport in which the concussion was incurred. The headform was then impacted at the velocity, mass, and location as defined in the video analysis [21, 22], producing linear and rotational acceleration loading curve signals. Three trials were conducted for each injury event with the signals from each trial analyzed in this research. The dependent variables used to quantify the impacts were peak resultant linear and rotational acceleration and maximum principal strain.

2.3 Signal processing

Once the event had been reconstructed in the laboratory, the x, y, and z axis linear and rotational acceleration signals were processed with CFC 60, 180, 600, and 1000 filters (Table 3). These filters were used because the CFC class filters are those that aremost commonly used in head and brain injury research according to the SAE J211 convention. In accordance with SAE J211, the CFC filter is a fourth-order Butterworth low pass with linear phase and special starting conditions. The filter sequence is described by the following difference equation (formula filter constants found in Table 4):

$$(3) \quad Y[t] = a_0X[t] + a_1X[t-1] + a_2X[t-2] + b_1Y[t-1] + b_2Y[t-2]$$

with $X[t]$ the input data sequence, $Y[t]$ the filtered output data sequence, a_0, a_1, a_2, b_1, b_2 the filter constants varying with CFC, and T the sampling rate in seconds. The difference equation describes a two-channel filter. For a fourchannel filter, the data of the two-channel filter was run twice: once forward and once backward, to prevent phase displacements. The filtered and unfiltered signals were then used as input for the UCDBTM to determine the effect of these filters on the maximum principal strain response. The linear and rotational acceleration time histories were input through the center of gravity of the brain to allow for the motion of the headform upon impact to be simulated using the UCDBTM [1]. Details on this method of application of acceleration time histories to determine brain stresses and strains using an FE model is described in Zhang et al.'s [1] and Yoganandan et al's [18] research.

Table 3 The formula filter constants calculations

$\omega_d = 2\pi \cdot CFC \cdot 2.0775$	$\omega_a = \dfrac{\sin \omega_d \frac{T}{2}}{\cos \omega_d \frac{T}{2}}$	$a_0 = \dfrac{\omega_a^2}{1+\sqrt{2}\omega_a+\omega_a^2}$	$b_2 = \dfrac{-1+\sqrt{2}\omega_a+\omega_a^2}{1+\sqrt{2}\omega_a+\omega_a^2}$
$a_1 = 2a_0$	$a_2 = a_0$	$b_1 = \dfrac{-2(\omega_a^2-1)}{1+\sqrt{2}\omega_a+\omega_a^2}$	

Table 4 CFC filter types and specifications.

	CFC 60	CFC 180	CFC 600	CFC 1000
3 dB limit frequency (Hz)	100	300	1000	1650
Stop damping (dB)	-30	-30	-40	-40
Sampling frequency	at least 600 Hz	at least 1800 Hz	at least 6 kHz	at least 10 kHz

Results

The effects of the different low-pass filters on the peak resultant linear and rotational acceleration loading curves and the resulting MPS measures from the UCDBTM can be found in

Tables 5 and 6. Graphical analyses of the effect of the filters on the signals are presented in Figs. 1 and 2.

Table 5 Mean peak (SD) results of the filtering effects for the American football reconstructions of four injury events.

Mechanism	Impact velocity (m/s)	CFC filter	Cutoff frequency (Hz)	Mean peak linear acceleration (g)	Avg % difference form unfiltered	Mean peak rotational acceleration (krad/s^2)	Avg % difference form unfiltered	Mean peak maximum principal strain	Avg % difference form unfiltered
Knee	7.89	100	100	106.4 (0.5)	31.8	8.9 (0.9)	45.2	0.544 (0.08)	15.3
		180	300	138.3 (4.5)	5.8	12.7 (2.1)	10.4	0.611 (0.08)	3.7
		600	1000	145.8 (7.7)	0.5	13.5 (2.5)	4.3	0.618 (0.08)	2.6
		1000	1650	145.7 (7.5)	0.6	13.5 (2.5)	4.3	0.612 (0.08)	3.5
		-	None	146.6 (8.2)	-	14.1 (2.7)	-	0.634 (0.05)	-
Shoulder	7.48	100	100	23.2 (0.3)	5.5	1.5 (0.2)	18.2	0.142 (0.05)	7.5
		180	300	24.3 (0.3)	0.8	1.7 (0.1)	5.7	0.138 (0.03)	10.3
		600	1000	24.3 (0.3)	0.8	1.7 (0.1)	5.7	0.139 (0.03)	9.6
		1000	1650	24.4 (0.3)	0.4	1.7 (0.1)	5.7	0.145 (0.04)	5.4
		-	None	24.5 (0.3)	-	1.8 (0.1)	-	0.153 (0.03)	-
Helmet	8.53	100	100	75.9 (2.3)	11.3	5.9 (0.2)	18.5	0.502 (0.004)	2.2
		180	300	84.1 (3.4)	1.1	6.8 (0.4)	4.3	0.514 (0.002)	-0.2
		600	1000	84.7 (4.0)	0.4	6.9 (0.5)	2.9	0.521 (0.010)	-1.5
		1000	1650	84.8 (3.9)	0.2	6.9 (0.5)	2.9	0.521 (0.015)	-1.5
		-	None	85.0 (4.0)	-	7.1 (0.5)	-	0.513 (0.013)	-
Fall	3.96	100	100	92.5 (3.1)	18.8	5.4 (0.05)	41.2	0.433 (0.01)	12.6
		180	300	109.7 (6.6)	1.8	7.9 (0.5)	3.7	0.482 (0.01)	1.8
		600	1000	110.7 (6.9)	0.9	8.1 (0.6)	1.2	0.482 (0.01)	1.8
		1000	1650	110.9 (7.1)	0.7	8.1 (0.6)	1.2	0.489 (0.003)	0.4
		-	None	111.7 (6.8)	-	8.2 (0.7)	-	0.491 (0.005)	-

Table 6 Mean peak (SD) results of the filtering effects for the ice hockey reconstructions of five injury events.

Mechanism	Impact velocity (m/s)	CFC filter	Cutoff frequency (Hz)	Mean peak linear acceleration (g)	Avg % difference form unfiltered	Mean peak rotational acceleration (krad/s^2)	Avg % difference form unfiltered	Mean peak maximum principal strain	Avg % difference form unfiltered
Elbow	7.3	100	100	31.5 (1.1)	8.2	3.0 (0.07)	12.5	0.231 (0.003)	1.7
		180	300	33.7 (0.8)	1.5	3.1 (0.03)	9.2	0.234 (0.001)	0.4
		600	1000	34.0 (0.8)	0.6	3.1 (0.03)	9.2	0.237 (0.01)	-0.8
		1000	1650	34.1 (0.8)	0.3	3.2 (0.03)	6.1	0.239 (0.007)	-1.7
		-	None	34.2 (0.9)	-	3.4 (0.06)	-	0.235 (0.004)	-
Fall	4.8	100	100	82.8 (8.6)	26.9	8.7 (0.2)	37.4	0.635 (0.02)	2.5
		180	300	102.7 (14.4)	5.5	11.6 (0.7)	9.1	0.670 (0.03)	-2.9
		600	1000	106.1 (14.1)	2.2	12.2 (0.2)	4.0	0.663 (0.02)	-1.8
		1000	1650	106.9 (13.1)	1.5	12.3 (0.1)	3.2	0.674 (0.03)	-3.5
		-	None	108.5 (11.2)	-	12.7 (0.1)	-	0.651 (0.01)	-
Boards	2.15	100	100	19.1 (0.1)	2.6	2.1 (0.008)	17.4	0.138 (0.03)	-2.9
		180	300	19.2 (0.1)	2.1	2.3 (0.01)	8.3	0.131 (0.02)	2.3

		600	1000	19.3 (0.1)	1.5	2.3 (0.02)	8.3	0.130 (0.01)	3.0
		1000	1650	19.3 (0.01)	1.5	2.4 (0.001)	4.1	0.136 (0.03)	-1.5
		-	None	19.6 (0.1)	-	2.5 (0.06)	-	0.134 (0.02)	-
Shoulder	7.9	100	100	27.2 (5.3)	10.8	2.3 (0.6)	29.6	0.225 (0.004)	-2.7
		180	300	29.0 (4.9)	4.4	2.6 (0.7)	17.5	0.251 (0.01)	-13.6
		600	1000	29.0 (4.9)	4.4	2.6 (0.7)	17.5	0.236 (0.01)	-7.5
		1000	1650	29.2 (5.2)	3.7	2.7 (0.8)	13.8	0.232 (0.01)	-5.8
		-	None	30.3 (6.5)	-	3.1 (0.7)	-	0.219 (0.02)	-
Puck	38.6	100	100	41.5 (2.7)	158.8	5.9 (1.3)	153.6	0.347 (0.05)	29.5
		180	300	125.7 (3.4)	96.6	12.5 (3.1)	113.0	0.438 (0.009)	6.4
		600	1000	247.4 (2.5)	37.6	20.7 (1.4)	74.0	0.460 (0.04)	1.5
		1000	1650	291.8 (1.3)	21.4	28.3 (1.6)	45.6	0.457 (0.03)	2.2
		-	None	361.9 (10.3)	-	45.0 (6.7)	-	0.467 (0.04)	-

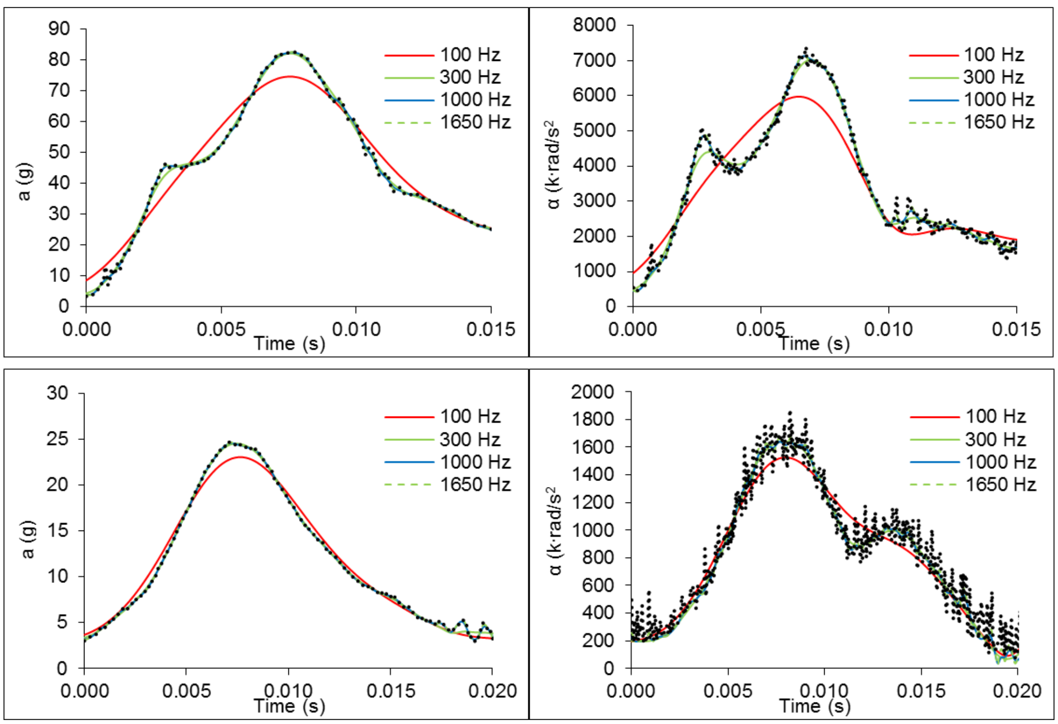

Figure 1. Graph of the effects of the filters for: (*top*) American football helmet to helmet impact; (*bottom*) shoulder to helmet impact. *Dotted line* is the unfiltered signal.

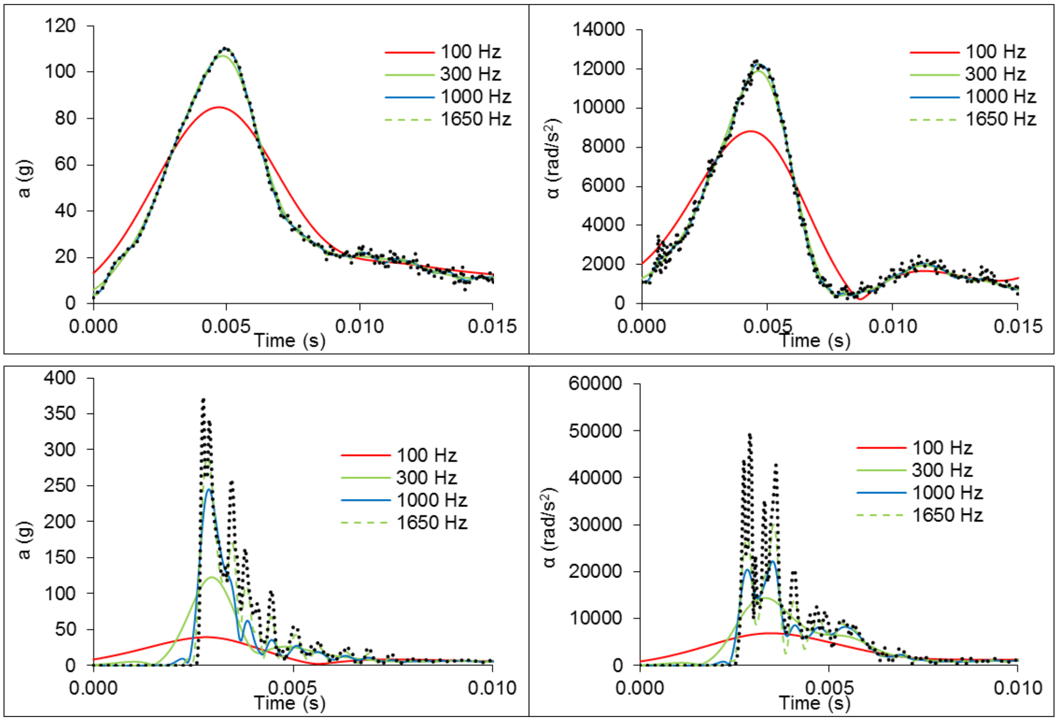

Figure 2. Graph of the effects of the filters for: (*top*) ice hockey fall to ice; (*bottom*) puck to helmet impact. *Dotted line* is the unfiltered signal.

4 Discussion

4.1 Effect of filtering on American football reconstruction signals

The events for the American football reconstructions included knee, shoulder, helmet, and falls, all impacting the head directly. The knee to head reconstruction had a raw data linear acceleration peak of 146.6 ± 8.2g. This peak value was not greatly affected by the filter, unless a filter with a cutoff of 300 Hz was used, which reduced the peak magnitude from 146.6 ± 8.2 to 138.3 ± 4.5g. A similar relationship was found for the peak mean rotational acceleration signal, where there was a larger drop in peak when the 300 Hz low-pass filter was applied. When examining the MPS results, the magnitude of strain was largely unaffected unless a low-pass filter with a 100 Hz cutoff was applied, resulting in approximately a 7 % change in strain results. The results of the FE model of the human brain are a reflection of the interaction of the acceleration loading curve with the material characteristics and definitions of the brain tissues. Since the output in strain is not directly dependent on peak magnitude acceleration, but rather a reflection of the entire motion over time, including slope, total duration of the event and other curve characteristics [19], it is likely that for these impacts the peak magnitude was not reduced enough by the effect of the filter to affect the peak strain. The remaining events follow a similar pattern in the effect of the filters on the response, with the exception of when the 100 Hz low-pass filter was applied, meaning filters with cutoffs above this may have little effect on response. There was a 2–6 % difference in the peak maximum principal strain from the raw signal in comparison to when a low-pass filter with a 300 Hz cutoff was applied. This result is reflected in Fig. 1, where the signals that have undergone 1650, 1000, or 300 Hz cutoffs closely follow the raw signal. The 100 Hz cutoff low-pass filter deviates from the original signal and has an effect on the resulting magnitudes of response. The effect the 100 Hz filter had on the peak magnitudes of response varied depending on the event, with peak linear acceleration magnitudes reduced by up to 40g for knee impacts, whereas for shoulder impacts the difference was only 1.5g. A similar relationship was found for the rotational acceleration (4000 vs. 100 rad/s2) for the same two event types. The effect on the MPS was less, with a reduction in peak magnitude of 4 % for the knee impacts and 2 % for the shoulder. The other two event types had similar reductions in peak magnitude of response in linear, rotational acceleration, and MPS, with the shorter duration, high magnitude events generally more affected by the filter with a low cutoff frequency. In comparison to the literature, the lower magnitude, longer duration impacts fell within the reductions in peak kinematics as described by Newman et al.; however, when the event generated a different signal type, such as the short duration high magnitude events of a knee or fall to the ground, the filters with a lower cutoff had a greater effect [14]. The results suggest that for signal durations of 10 ms or more, low-pass filters with a cutoff frequency above 300 Hz would have minimal effect on the kinematic and finite element responses. For the shorter duration events, such as those below 10 ms, the results indicate that the choice of low-pass filter will have more of an effect on the responses.

4.2 Effect of filtering on ice hockey head impact reconstruction signals

There were five events that were reconstructed for ice hockey including impacts involving elbows, falls, impact to the boards, shoulder, and hit by a puck. These represent the most common events resulting in concussive injury in the sport of ice hockey [20]. The elbow, boards, and falling data showed minimal reduction in peak magnitude of linear and rotational acceleration when a 100 Hz cutoff frequency filter was used (1–4g; 100–700 rad/s2), meaning filters with cutoff frequency above 100 Hz would have little effect on responses. Interestingly, these reductions in peak linear and rotational accelerations (up to 4g linear, and 700 rad/s2 rotational) did not greatly affect the MPS peak result, with a difference of approximately 1.5 % strain in these two events. The effect of the different filters for shoulder to head impacts was small for the peak linear (2g) and rotational acceleration (700 rad/s2), but had more of an effect on the MPS with a difference of approximately 2 % (Figs. 1, 2; Table 6). Among all the ice hockey reconstructions, the puck impacts were the signals that were the most affected by the application of the filters. The peak linear (reduction up to 320g) and rotational accelerations (reduction up to 39,100 rad/s2) were greatly reduced depending on the filter used, and this was also reflected in the MPS magnitudes (up to 12 %). Of all the reconstructions, this type of impact was the one with the lowest duration, and with the highest peak magnitudes in both linear and rotational acceleration (Fig. 2). It is likely that the filters used have a greater effect on short duration signals such as the 5 ms durations found for puck impacts. For these types of signals, even low-pass filters with cutoffs as high as 1650 Hz can have an effect and may even be necessary as a result of the high-frequency noise which seems evident in the raw unfiltered signal (Fig. 2). The low-pass filters with cutoffs of 300 Hz and above had little effect on the signals with durations above 10 ms, suggesting that these types of signals may be less sensitive to this type of signal conditioning.

Conclusion

The purpose of this research was to investigate the effects of processing linear and rotational acceleration signals with a variety of low-pass filters and how that may affect finite element modeling output, in this case measured in maximum principal strain. In many cases, particularly when the duration of the signal was above 10 ms, the effect of using a filter with a cutoff frequency of 300 Hz or higher had little effect on the peak MPS. This research demonstrates that using a 1000 Hz cutoff for direct head impact reconstructions using a Hybrid III headform may be suitable for events that are longer in durations (over 10 ms). In fact, using a 300 Hz cutoff frequency seems to have a minimal effect on the results. Overall, when examining these results, short duration (less than 10 ms) signals with high magnitudes are most likely to be affected by low-pass filters with cutoffs similar to those used in this research, with the lowest cutoff frequencies (100 Hz) having the largest effect.

Compliance with ethical standards

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