Dynamic Response under Short Acting Forces

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Abstract :

ACCELERATIONS experienced by human subjects during the application of short acting forces (eg : ejections from aircraft) are significantly different from those applied to the system because of the viscoelastic properties of the man – seat pack assembly. Dynamic Response Index which is being currently used in the evaluation of ejection accelerations is discussed alongwith the percentage spinal fractures obtained in ejections.

The paper also describes a novel method of determining the dynamic overshoot of accelerations during the ejection boost based on the frequency dependent mechanical impedence of human subjects. Experimental support for this concept is furnished from ejection trials using Anthropomorphic Dummy.

Introduction :

The vertebral column of a human body is structurally comprised of rigid bony vertebrae, cartilagenous intervertebral discs with ligament attachments. Thus this forms an elastic system capable of a dynamic response. The inorganic and organic constituents of this structure offer high compressive tensile stiffness. Flexion compression and expansion which are the properties of this structure mainly depend upon its elasticity. This property in turn provides a man with postural stability, lateral and rotational bending. The convex – concave arrangement of various intervertebral discs of the structure contributes greatly to the overall spinal elasticity and thus dynamically maintain the centre of gravity in a man.

The rate of onset of acceleration[1] during an initial phase of ejection, together with the associated elastic properties of the human body and the structure can cause an acceleration overshoot. This sometimes increases the inertial loading on the spine resulting in spinal compression and subsequent fracture.

In the ejection seat assembly, when a man is ejected out, the elastic qualities of the seat pack and cushions will also contribute to the dynamic overshoot of acceleration[2]. The forces generated by the ejection cartridges and applied to the ejection seat get modified significantly as a result of internal dynamics of the man – seat system. The safety criteria, for the ejection forces have to be based on the dynamic response of the human subject. Von Gierke[3] had suggested the use of dynamic response index, with the ejection type of forces. This concept is being used in the USAF and is now being adopted by the RAF and NATO countries[4].

There is a large percentage of vertebral injuries during ejections ranging upto 40% in the various Air Forces of the world using different types of ejection seats and cartridges. Critical control of the 'G' profile experienced by the ejected aircrew's is essential in minimising the injury rates and magnitudes of the injury, and thereby preventing permanent incapacitation, and long periods of grounding of expensively trained and experienced aircrew.

Dynamic Response Index and its Determination:

The dynamic response index which is being currently used in US and NATO countries

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is based on a single mass-spring model with a damping element:

$$\frac{d^2\delta}{dt^2} + 2KWn\frac{d\delta}{dt} + W^2n\delta = \frac{d^2z}{dt^2} \qquad (1)$$

where δ is the compression of the spring in feet, K, the damping ratio of the model, Wn is its undamped natural frequency in radians/sec and $\frac{d^2z}{dt^2}$ is the Z axis acceleration.

The DRI[2] is given as:

$$DRI = \frac{W^2n\ \delta_{max}}{g}$$

The values of K and Wn for a model equivalent to the human body have been estimated to be:

$$K = 0.224 \text{ and}$$

$$Wn = 52.9 \text{ rad/sec}$$

Applying these numerical values:

$$\frac{d^2\delta}{dt^2} + 23.7\frac{d\delta}{dt} + 2798\,\delta = \frac{d^2z}{dt^2} \qquad (2)$$

and

$$DRI = 86.9\ \ \delta\ \max \qquad (3)$$

In actual practice DRI is obtained from δ max estimated from the G profile by a computer.

DRI and Safety Criteria:

Table 1 gives the DRI values against percentage risk of injury based on ejections in the USAF.

TABLE 1

DRI and Injury Probability

Simple forces		Complex forces	
DRI	Risk of injury	DRI	Risk of injury
18.0	5%	17.0	5–20%
20.4	5–20%	19.0	20–50%
23	50%	22.0	50%

The DRI values obtained from δ max determined by computer from the G-profile are compared against the injury potential given in Table 1.

A study has been made on the UK data of ejections for the following aircrafts — Gnat, Hunter, Javelin and Lightning. The actual injury rates were calculated from the number of cases of spinal fractures in relation to the total number of non-fatal accidents. The results are summarised in Table 2.

TABLE 2

DRI and Injury potential for different Aircraft and Seats

Aircraft & seat	Ejection data		Injury rate	DRI based on G values: for seat	DRI based on G values: for subjects
Gnat Folland seat	1. Total:	26		20	21
	2. Fatal:	4	27%	19	19
	3. Vertebral			—	20
	fracture:	6		—	21
				—	20
					21.2
Hunter Martin Baker Mk D seat	1. Total:	13		·22.3	22.6
	2. Fatal:	1	50%	24.3	23.8
				23.9	23.6
	3. Vertebral fracture:	6		23.6	24.5
Lightning Martin Baker Mk 4 seat	1. Total:	32		17.4	18.4
	2. Fatal:	7	40%	18.6	19.5
	3. Vertebral fracture:	10		19.4	18.4

Graphical representation of the DRI against injury potential incorporating the DRI ranges for different aircrafts and seats is given in Fig. 1. Apparently, when DRI values are higher as in Javelin and Hunter, the injury rates are lower than those predicted by DRI. When DRI values are lower as in Gnat and Lightning, the injury rates are higher than those predicted by the DRI line. This shows the limitations of the current concept of DRI.

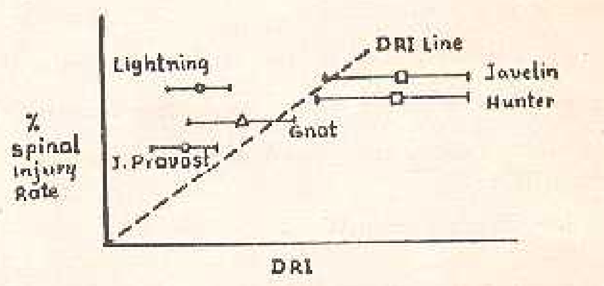

Fig. 1

Limitations of Dynamic Response Index :

The concepts of dynamic response index have got a number of limitations :

(*i*) The dynamic equivalent of a single mass-spring is an over simplified approach.

(*ii*) The spinal injury rates are the ratios of compression fractures to non-fatal ejections with various types of seats, harness system and survival packs, and are not generated from actual data from internal organ damage.

(*iii*) Analysis of the DRI obtained for a number of aircraft in RAF and comparison with injury percentage reveal that with low DRI values, the injury rates are higher ; and when DRI values are higher then, the injury rates becomes lower than the predicted values.

Mechanical Impedence and Force Amplification :

In the present study an attempt is made to explain the acceleration overshoot on the basis of the mechanical impedence of the human body. The mechanical impedence of the system on analogy with the electrical equivalent can be expressed as the ratio of the transmitted force to the velocity of that point where the force is transmitted.

$$Z = \frac{F}{\dot{X}}$$ – where Z is the impedence

F is the force and $\dot{X}$ the velocity.

On the basis of the dynamic equivalent of a single mass-spring-damping model the relation of $|Z|$ with the elements of the system can be expressed as follows :

$$|Z| = m.w. \sqrt{\frac{\delta^2 n^2 + 1}{(1-n^2)^2 + n^2\delta^2}} \qquad (4)$$

Where m is the mass, $\delta = \frac{D}{W_o m}$ where D is the damping constant, W_o is the undamped natural frequency and $n = \frac{W}{W_o}$ where W is the frequency of forced vibration, the impedence of a pure mass will then be :

$$Z \text{ mass} = mW$$

The difference in the impedence values between mass (seat) and the model (hip of subject/dummy) can be obtained from the above relations. Figure 2 gives the impedence values for a seated human subject for different frequencies of forced vibrations.

The deviation of the impedence of the seated subject for any frequency from that of a rigid mass can be obtained from Fig. 2. Peak values of acceleration obtained from a rigid structure (*eg.: data* from a seat mounted accelerometer) when modified for the impedence deviation will give the peak acceleration experienced by the subject.

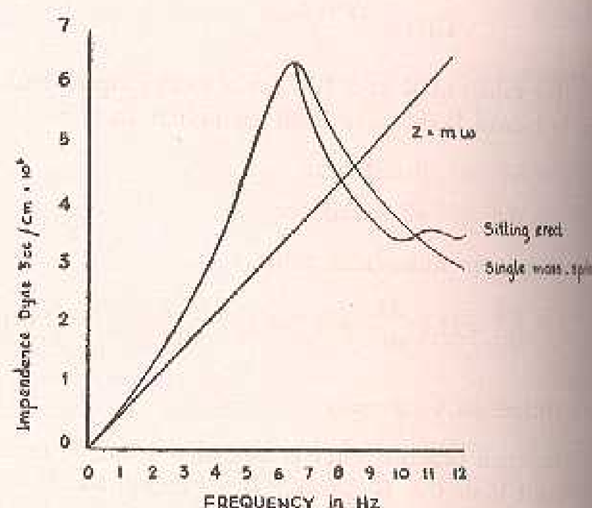

FIG- 2

Experimental Trials :

Experimental validation for this approach is provided by the G-profiles collected during ejections using a test rig. Anthropomorphic dummy developed at the Institute of Aviation Medicine was seated firmly in an ejection seat which was mounted on an ejection tower. A rigid survival pack with a crushable cushion was used between the seat pan and the dummy. The dummy was instrumented with two accelerometers one in its hip joint while the other was fixed on the seat structure, to give a picture of relative accelerations on hip and seat simultaneously during an ejection force.

G-profiles were recorded on an ultra violet Viscorder with paper set at a speed of 800 mm/sec, with time markings at 100 m. sec.

Results and Discussion :

The records of G-profiles obtained from the ultra violet recorder were traced for analysis. These

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records are given in Fig. 3 (a) and 3 (b). From the records peak values of G and rate of onset for the seat and hip of the dummy are calculated. The results are given in Table 3.

TABLE 3

Peak G and Rate on Onset

Cartridge No.	Peak G at seat	Rate of onset at seat level	Peak G at hip	Rate of onset at hip
1	14.5	185.5	18.6	247.0
2	14.3	173.0	18.1	288.0
3	13.9	161.0	16.4	235.0
4	12.5	161.0	16.5	252.0
5	12.8	161.0	16.0	264.5
6	13.8	167.0	17.6	188.0
7	12.0	148.5	14.5	200.0
8	13.1	173.0	16.8	206.0
9	13.8	167.0	15.9	270.0
10	13.0	192.0	16.2	253.0

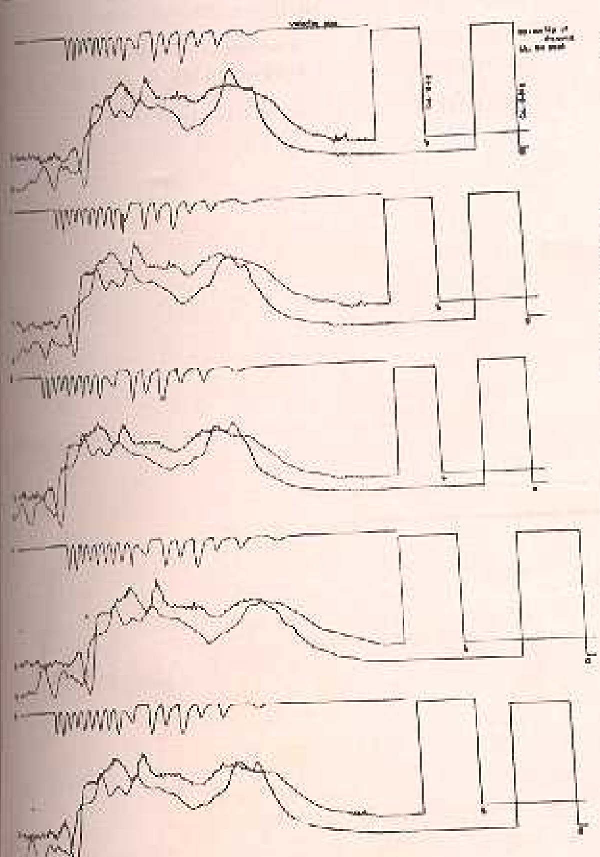

Fig. 3(a)

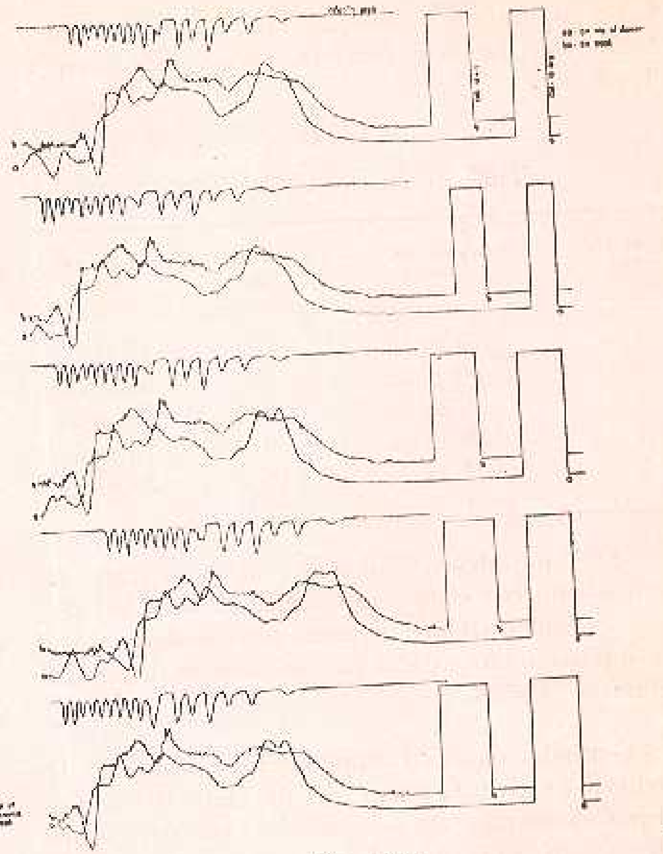

Fig. 3(b)

The difference in the G values at hip level and seat level from the experimental firings can be obtained from Table 3. The average difference in the peak G values is 27%; hip accelerations being higher. The rate of onset, which is more critical a factor in injury analysis, shows an average difference of 42%.

An analysis of the onset of acceleration on the G-profiles (Fig. 3 (a) and 3 (b)) gives the following salient points:

(*i*) An initial slow onset for approximately 10 m.secs with extremely small G values.

(*ii*) An onset approximating to a straight line.

(*iii*) An onset approximating to a sine wave at the end of the straight line onset.

Treating the latter half of straight line onset and the curved onset at the end as approximating to a sine wave application, the duration of onset can be worked out and the equivalent frequency of the force application estimated. The values thus calculated for the ten different curves are given in Table 4.

TABLE 4

Time of onset for Sine Wave

Cartridge No.	Time of onset	Cartridge No.	Time of onset
1	75 m.sec	6	75 m.sec
2	80 m.sec	7	80 m.sec
3	75 m.sec	8	70 m.sec
4	80 m.sec	9	75 m.sec
5	75 m.sec	10	75 m.sec

The impedence difference in percentage for human subjects compared to a rigid mass is 35% (Fig. 2) and therefore accounts for the deviation of the hip accelerations and rate of onset from the seat values.

G-profiles obtained from ejection seat can be modified to give G-values at the hip level of a subject knowing the impedence of the seated subject. In applying this novel approach, the equivalent natural frequency of the applied force is first determined approximating the latter portion of the onset G-profile to be a sine wave. Impedence for seated subject for this frequency is determined and its deviation from the impedence of a rigid mass is estimated.

In the evaluation of ejection seat cartridges acceleration profiles obtained from seat mounted accelerometers can be used provided a correction is applied based on the impedence approach. The corrected value will give the acceleration at the hip level. Accelerations at the hip level can then be compared against human tolerance values for accepting a particular type of cartridge.

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