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Theoretical and experimental model to describe the injection of a polymethylmethacrylate cement into a porous structure

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Abstract

A theoretical approach was used to determine the distribution of a poly(methylmethacrylate) cement after its injection into a porous structure. The predictions of the model were then compared to experimental results obtained by injecting a polymethylmethacrylate cement into an open-porous ceramic filter. The goal was to define a model that could predict what factors affect the risk of cement extravasation and hence how the risk of cement extravasation can be minimized. The calculations were based on two important rheological laws: the law of Hagen–Poiseuille and the law of Darcy. The law of Hagen–Poiseuille describes the flow of a fluid in a cylindrical tube. The law of Darcy describes the flow of a fluid through a porous media. The model predicted that the extravasation risk was decreased when the cement viscosity, the bone pore size, the bone permeability and the bone porosity were increased, and when the diameter of the extravasation path and the viscosity of the marrow were decreased. Experimentally, the effect of the marrow viscosity and extravasation path could be evidenced. Therefore, the model was believed to be an adequate approximation of the experimental behavior. In conclusion, the experimental results demonstrated that the model was adequate and that the best practical way to decrease the risk of extravasation is to increase the cement viscosity.

Keywords: Polymethylmethacrylate; Cement; Extravasation; Viscosity; Model; Injection

1. Introduction

Due to population ageing, osteoporosis is becoming one of the largest medical problems [1]. Osteoporosis leads to an enormous increase of the incidence of bone fracture. Additionally, the fixation of fractures in osteoporotic bone can be extremely difficult due to the limited anchoring force of bone screws. Considering this problem, surgeons have proposed to reinforce or stabilize osteoporotic bone with polymethylmethacrylate (PMMA) bone cement. With the use of PMMA, it is possible to achieve an adequate anchorage of the implant or to prevent further collapse, for example, in vertebral bodies. PMMA cements hardens via a very exothermic reaction, which may lead to bone necrosis. Additionally, the monomer of PMMA is toxic. Despite these drawbacks, PMMA cements have been increasingly used because the outcome expressed in terms of fracture stability and pain relief is very good. So far, a large number of studies have been published on the topic, ranging from the clinical use of PMMA cement [2–5], to the effects of the exothermic setting reaction [6], via the selection of an adequate material [4,7,8], or the mechanical effects of bone augmentation [8–10]. Surprisingly, very few studies have been devoted to the rheological properties of the cement [11–13], even though the cement viscosity plays a key role in determining the force required to inject the cement, the distribution of the cement in bone, and the occurrence of cement extravasation [3,14,15]. More importantly, there is to our knowledge no study linking the degree of bone filling with the viscosity of the cement. In most studies, the conditions for cement injection are hardly described. These conditions include the time elapsed between the start of mixing and the injection, the injection rate, the needle size, or the syringe size. Moreover, surgeons tend

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Nomenclature

ΔP_n	pressure drop in a fluid flowing through a needle (Pa).
dP	increment of pressure (Pa).
ΔP_r	pressure drop between the tip of the needle and a distance r of the injection point (Pa).
ΔP_∞	pressure drop between the tip of the injection needle and the external cement surface at infinite distance (Pa).
ΔP_a	total pressure required to augment bone with the cement (Pa).
$\Delta P_{a,c}$	pressure required to inject the cement into bone (Pa).
$\Delta P_{a,m}$	pressure required to extrude the marrow out of the bone (Pa).
ΔP_e	total pressure required to extravasate the cement (Pa).
$\Delta P_{e,c}$	pressure required to inject the cement through the path of least resistance (Pa).
$\Delta P_{e,m}$	pressure required to extrude the marrow through the path of least resistance (Pa).
ΔP_p	pressure drop in a cylindrical pore (Pa).
Q_f	flow rate of the fluid flowing through a needle (m^3/s).
Q_c	flow rate of the cement injected into bone (m^3/s).
Q_m	flow rate of the marrow flowing out of bone (m^3/s).
q_p	flow rate of a fluid injected into a cylindrical pore (m^3/s).
Q_p	global flow rate of a fluid injected into n_p cylindrical pores (m^3/s).
v	average velocity of the fluid flowing through a porous matrix (m/s).
μ_f	fluid viscosity (Pa s).
μ_c	cement viscosity (Pa s).
μ_m	marrow viscosity (Pa s).
dx	increment of distance (m).
dr	increment of radius (m).
r	radial distance to the injection point (m).
R_0	radius of the cavity at the injection point (m).
R_v	radius of the bone sphere (= L_e in our model) (m).
r_c	radius of the sphere filled with bone cement (m).
D_p	diameter of the cylindrical pores (m).
d	distance between the injection point and the center of the extravasation path (Fig. 4).
K	matrix permeability (m^2).
S_m	surface area of the matrix (perpendicular to the cylindrical pores) (m^2).
p	matrix porosity (dimensionless).
D_n	needle diameter (m).
D_e	diameter of the path of least resistance (m).
L_n	needle length (m).
L_e	length of the path of least resistance (= R_v) (m).
L_c	length of the narrow path filled with cement (m).
L_m	length of the narrow path filled with marrow (m).
L_p	length of the cylindrical pores (m).
n_p	number of cylindrical pores (dimensionless).
τ	matrix tortuosity (dimensionless).
t_e	time to fill the extravasation path with cement (s).
t_a	time to augment bone with cement (s).
λ	extravasation risk factor (dimensionless).

to use a different powder-to-monomer ratio than the value adviced by the cement producer to prolong the injection period [5,16]. This can lead to large changes of the physical properties of the cement, in particular the rheological and mechanical properties [16]. Furthermore, a contrasting agent is typically added to improve the visualization of the cement during injection and hence decrease the extravasation risk [2,5,9,10]. Examples of contrasting agents used in vertebroplasty are barium sulfate powder [9], tungsten powder [5], tantalum powder [2–4], and an iodine solution [8]. Finally, all results depend on the structure of the patient bone which of course vary a lot from one patient to the other. Therefore, it is difficult to determine adequate conditions for bone augmentation based on clinical or cadaveric studies.

The goal of the present study is to understand what flow characteristics a cement must have to be a good bone augmentation material. For that purpose, a theoretical approach is used. Two important rheological laws are considered: the law of Hagen–Poiseuille and the law of Darcy. A particular attention is devoted to the changes of pressure during cement injection and possible risks of cement extravasation. In the second part of the study, experiments are performed and compared with theoretical predictions.

2. Theoretical

The goal of this section is to predict theoretically the behavior of the cement paste during its injection into a porous matrix. A particular attention is paid to the effects of injection parameters (flow rate, cement viscosity, matrix geometry) on the injection force and the occurrence of extravasation. The calculations are

based on the law of Hagen–Poiseuille and the law of Darcy. The law of Hagen–Poiseuille describes the flow of a fluid in a cylindrical tube. The law of Darcy describes the flow of a fluid through a porous media. These two laws can be combined to simulate the behavior of a cement paste injected through a needle into a porous matrix, e.g. an osteoporotic cancellous bone.

2.1. Approach

The Hagen–Poiseuille relationship relates the pressure drop, ΔP_n, in a fluid flowing in a needle to the flow rate of the fluid, Q_f.

$$Q_f = -\frac{\Delta P_n \pi D_n^4}{128 \mu_f L_n}, \tag{1}$$

where μ_f is the viscosity of the fluid, D_n is the diameter of the needle, and L_n is the length of the needle. This relationship is only valid when the flow is laminar and when the fluid is Newtonian, i.e. when the fluid viscosity is independent of the shear rate. In the present case, the flow is laminar, but the fluid is non-Newtonian [12]. However, this relationship will be considered to be valid in a first approach.

The law of Darcy describes the pressure drop in a fluid flowing through a porous media:

$$dP = -\frac{v \mu_f}{K} dx, \tag{2}$$

where dP is an increment of pressure, dx is an increment of distance, v is the average velocity of the fluid, and K is the permeability of the porous media. If a Newtonian fluid is injected into the center of an open-porous and isotropic filter, the fluid is expected to spread spherically, so the Darcy law should be integrated in spherical coordinates. Assuming no flow distortion at the injection point, the differential equation becomes

$$dP = -\frac{v \mu_f}{K} dr, \tag{3}$$

where dr is an increment of radius. The average speed of the fluid at a distance r of the injection point is a function of the flow rate and the matrix porosity, p

$$v = \frac{dr}{dt} = \frac{Q_f}{4\pi r^2 p}. \tag{4}$$

Eq. (4) can be combined with Eq. (3):

$$dP = -\frac{Q_f \mu_f}{4\pi p K r^2} dr. \tag{5}$$

This equation can be integrated

$$\Delta P_r = -\frac{Q_f \mu_f}{4\pi p K}\left(\frac{1}{R_0} - \frac{1}{r}\right), \tag{6}$$

where R_0 is the radius of the cavity at the point of injection and ΔP_r is the pressure drop between the tip of the needle and a point at a distance r of the injection point ($r \geqslant R_0$). Interestingly, the absolute pressure drop is limited to a maximum of ΔP_∞, corresponding to the pressure drop between the injection point and the fluid surface at infinite distance

$$\Delta P_\infty = -\frac{Q_f \mu_f}{4\pi p K} \frac{1}{R_0}. \tag{7}$$

This relationship shows that the pressure required to inject a fluid into a porous matrix depends very strongly on the porosity, p, and the geometry of the matrix permeability, K. Using Eq. (7), Eq. (6) can be written

$$\Delta P_r = \Delta P_\infty \left(1 - \frac{R_0}{r}\right). \tag{8}$$

When a cement is injected into an osteoporotic vertebral body, there is often extravasation of the cement. The risk of extravasation can be as high as 40% [2]. When extravasation occurs, the cement does not spread spherically into the bone, but follows a path of least resistance, for example, a blood vessel or a fracture gap. In other words, the pressure required to inject the cement into the path of least resistance, called here "extravasation pressure", ΔP_e, is smaller than the pressure required to expand the cement spherically, called here "augmentation pressure", ΔP_a. These two pressures can be calculated and compared to determine which parameters increase the risk of extravasation. For that purpose, Eqs. (1) and (7) must be modified to take into account the presence of cement and marrow in the bony structure.

It is assumed that the path of least resistance (extravasation path) is cylindrical, has a diameter, D_e, and a length, L_e (Fig. 1; therefore, $L_e = R_v$). Additionally, it is assumed that the cement and the marrow have a viscosity, μ_c and μ_m, respectively. The cement can only extravasate if it pushes the marrow out of the way. Therefore, the total pressure required to inject the

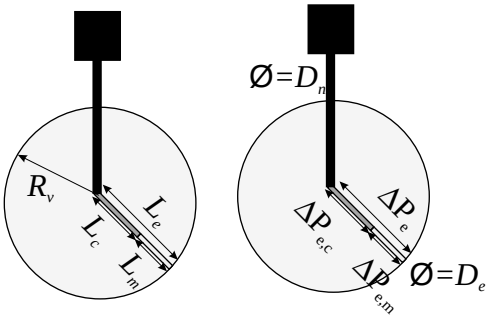

Fig. 1. Model for cement extravasation. The cement goes from the syringe into the needle (in black) and into the path of least resistance (length L_e; pressure drop ΔP_e). The parts of the path filled with cement and marrow have a length L_c and L_m, respectively. The pressure drop corresponding to these two segments have the values $\Delta P_{e,c}$ and $\Delta P_{e,m}$, respectively. The diameter of the needle and the path are D_n and D_e, respectively. The sphere represents an ideal bone volume that has to be filled with cement.

cement through this path, ΔP_e, is expressed by a combination of two equations, one describing the pressure required to inject the cement into the path, $\Delta P_{e,c}$, and one describing the pressure required to extrude the marrow out of the path, $\Delta P_{e,m}$

$$\Delta P_e = \Delta P_{e,c} + \Delta P_{e,m} = -\frac{128\mu_c L_c Q_c}{\pi D_e^4} - \frac{128\mu_m L_m Q_m}{\pi D_e^4}, \quad (9)$$

where Q_c and Q_m are the flow rate of cement and marrow, respectively. As the mass balance is constant during injection, $Q_c = Q_m = Q_f$. L_c and L_m are the lengths of the narrow path filled with cement and marrow, respectively (Fig. 1). These two lengths are linked to the length of the narrow path, $L_e(= R_v)$

$$L_e = L_c + L_m. \quad (10)$$

Additionally, these lengths are related to the injected cement volume

$$L_c = R_0 + \frac{4Q_c}{\pi D_e^2} t. \quad (11)$$

Here, we assume that the initial length ($t = 0$) is R_0, the radius of the injection cavity. After a time, t_e, the whole length of path of least resistance is filled with cement

$$t_e = \frac{\pi D_e^2}{4Q_c}(L_e - R_0). \quad (12)$$

Eqs. (9)–(11) can be combined to obtain the total pressure required to inject the cement into the extravasation path

$$\Delta P_e = -\frac{128Q_c}{\pi D_e^4}\left(\mu_c L_c + \mu_m\left(L_e - \frac{4Q_c}{\pi D_e^2} t\right)\right) \text{ if } t < t_e, \quad (13)$$

$$\Delta P_e = -\frac{128Q_c}{\pi D_e^4}(\mu_c L_e) \text{ if } t \geq t_e. \quad (14)$$

Here it is assumed that the extravasation pressure is constant when $t > t_e$, i.e. when the cement has completely filled the extravasation path.

The pressure required to augment bone with the cement is a function of two pressures (Fig. 2): the pressure required to inject the cement into the bone matrix, $\Delta P_{a,c}$, and the pressure required to extrude the marrow out of bone, $\Delta P_{a,m}$

$$\Delta P_a = \Delta P_{a,c} + \Delta P_{a,m}, \quad (15)$$

where $\Delta P_{a,c}$ and $\Delta P_{a,m}$ are (Eq. (6))

$$\Delta P_{a,c} = -\frac{Q_c \mu_c}{4\pi p K}\left(\frac{1}{R_0} - \frac{1}{r_c}\right), \quad (16)$$

$$\Delta P_{a,m} = -\frac{Q_m \mu_m}{4\pi p K}\left(\frac{1}{r_c} - \frac{1}{R_v}\right). \quad (17)$$

Here, it is assumed that the cement is injected into a bone sphere of radius R_v. The spherical volume filled with the cement has a radius r_c. The latter radius has a minimum value of R_0 (radius of the injection cavity at time zero) and is a function of time:

$$r_c = \left(R_0^3 + \frac{3Q_c}{4\pi p} t\right)^{1/2}. \quad (18)$$

After a certain time, t_a, the whole bone is filled with cement

$$t_a = \left(\frac{4\pi p\left(R_v^3 - R_0^3\right)}{3Q_c}\right). \quad (19)$$

Eqs. (15)–(19) can be combined to calculate the total pressure needed to augment bone

$$\Delta P_a = -\frac{Q_c}{4\pi p K}\left[\mu_c\left(\frac{1}{R_0} - \frac{1}{r_c}\right) + \mu_m\left(\frac{1}{r_c} - \frac{1}{R_v}\right)\right] \text{ if } t < t_a, \quad (20)$$

$$\Delta P_a = -\frac{Q_c \mu_c}{4\pi p K}\left(\frac{1}{R_0} - \frac{1}{R_v}\right) \text{ if } t \geq t_a. \quad (21)$$

Here again, it is assumed that the augmentation pressure is constant when $t \geq t_a$, i.e. when the cement has completely filled bone. The ratio between the augmentation pressure and the extravasation pressure is

$$\lambda = \frac{\Delta P_a}{\Delta P_e} = \frac{D_e^4}{512 p K} f\left(\mu_c, \mu_m, t, R_0, L_e\right), \quad (22)$$

where $f(\mu_c, \mu_m, t, R_0, L_e)$ is a function of the parameters within brackets and λ is the risk factor for extravasation. It is important to underline that in our model $L_e = R_v$. Moreover, the parameters L_c and L_m do not appear in Eq. (22) because they can be expressed as a function of L_e and the flow conditions (Eqs. (10) and (11)). When a cement is injected into a porous matrix, extravasation

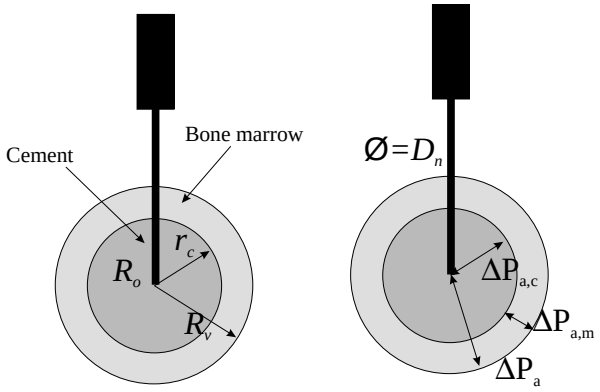

Fig. 2. Model for bone augmentation. The cement (in white) flows from the needle (in black, diameter D_n, assumed to be equal to R_0) into the center of spherical bone (radius R_v). The central part with radius "r_c" is filled with cement. The external part is filled with marrow. The black part represents a syringe coupled to a needle of diameter R_0. The pressure drop in the cement and the marrow are equal to $\Delta P_{a,c}$ and $\Delta P_{a,m}$ (total ΔP_a). The sphere represents an ideal bone volume that has to be filled with cement.

occurs when the extravasation pressure, ΔP_e, is smaller than the augmentation pressure, ΔP_a, i.e. when λ is larger than 1.

2.2. Analysis

Eq. (22) shows that the risk of extravasation increases when the matrix porosity, p, decreases, when the permeability, K, decreases, and when the diameter of the extravasation path, D_e, increases. Interestingly, the permeability decreases with a decrease of pore size (Appendix, Eq. (A.7)). Therefore, a decrease of pore size is expected to increase the risk of extravasation. The effect of the injection size, R_0, is not as trivial as the effect of p, K, and D_e. However, a closer look at Eqs. (13), (14), (21) and (22) shows that R_0 only affects the augmentation pressure: an augmentation of R_0 leads to a decrease of the augmentation pressure and hence a decrease of the extravasation risk. R_0 can be estimated to correspond to the pore size of bone. Therefore and once again, an increase of the bone pore size reduces the risk of extravasation.

To enlighten the effect of μ_c, μ_m, and t on the risk of extravasation, the function $f(\mu_c, \mu_m, t, R_0, R_v)$ was calculated for different viscosity ratios and various injected amounts (Fig. 3). It should be mentioned that the results represented in Fig. 3 are only qualitative and not quantitative. The λ values correspond one specific set of values of the different parameters (e.g. for one specific R_0). However, the general trends are valid. The calculations show that λ varies a lot with the injected volume (Fig. 3). After a short initial phase, corresponding to the zone where the injected volume is smaller or equal to the volume of the extravasation path (i.e. for $t \leqslant t_e$—Eq. (12)), there is a progressive change of the extravasation risk. The latter change is only determined by the ratio between the viscosity of the marrow and that of the cement: a low ratio leads to a constant increase of the risk, whereas a large value leads to a constant decrease of the risk. However, the latter changes are moderate compared with the change of the risk due to a variation of the ratio between marrow- and cement viscosity: a low ratio leads to a low risk, whereas a high ratio leads to a high risk. Therefore, the cement viscosity should be maximized to minimize the extravasation risk.

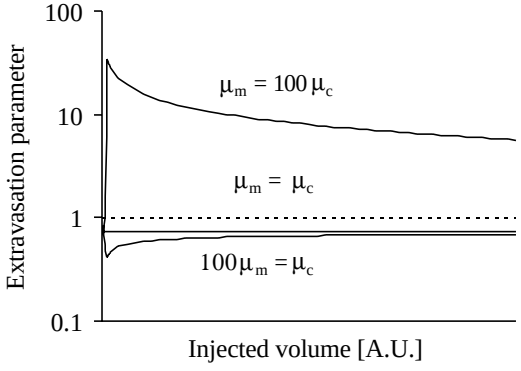

Fig. 3. Effect of viscosity and injected volume on the extravasation risk. When the extravasation factor has a value larger than one (i.e. above the dotted line), extravasation occurs. Two cases were considered: (a) The viscosity of the marrow, μ_m, was assumed to be 1 Pa s and the viscosity of the cement, μ_c, was varied. (b) The viscosity of the cement, μ_c, was assumed to be 1 Pa s and the viscosity of the marrow, μ_m, was varied. Both cases gave the same results. As a result, only case (a) was represented here.

One parameter that also affects the risk of extravasation is the size of the bone sphere, R_v. The calculations performed with the present model show that an increase of R_v tend to decrease the risk of extravasation. However, the effect is small.

3. Experimental

The PMMA cement. "Palacos LV-40 + gentamicinum" (Essex Chemie AG, Luzern, Switzerland) was used here. "Palacos LV-40" is a so-called low-viscosity bone cement [13], i.e. it has an extended low-viscosity phase. At the experimental temperature used in this study (22 ± 1°C), the cement viscosity is low for about 3.5 min after the start of mixing [14]. In the following 2.5 min, the viscosity quickly increases and the dough becomes warm. The cement hardens after approximately 7–8 min.

Each cement package contained 2 × 41.8 g powder and 2 × 20 ml monomer solution. Each package was divided into 8 equal parts (10.45 g powder, 4.7 ml monomer solution). Each part was mixed with a spatula in a 20 ml syringe (Discardit II, Becton Dickinson, Basle, Switzerland) for 40 s and injected into one 50 ml luer-lock syringe (Product No. 300865, Becton Dickinson, Basle, Switzerland). A needle (see Table 1) was screwed onto the 50 ml syringe and the force required to inject the cement into an open-porous ceramic filter was recorded. The ceramic filter had a cubic shape with a side-length of 5 cm. A 2.5 cm-long cylindrical hole was drilled into the block. The hole had the same diameter as the needle, and was centered in the middle of the top face, perpendicular to it. Therefore, the tip of the hole was exactly in the middle of the block. A second cylindrical hole was drilled perpendicularly to the first cylindrical hole. This hole was centered 15 mm above the bottom face, parallel to it (Fig. 4).

The experiments were done according to a factorial design of experiment $2^{(6-1)}$ (one repeat). The six parameters tested here were (Table 1): (A) filter filling (see hereafter); (B) needle gauge; (C) time elapsed after cement mixing before injecting the cement; (D) injection speed; (E) filter pore diameter; and (F) diameter of the side-hole. The first fluid used to fill the filters was

Table 1
Factors tested in the factorial design of experiments $2^{(6-1)}$ with one repeat

Factor	Low level	High level
(A) Filter filling with butter	No	Yes
(B) Needle diameter	Gauge 7[a]	Gauge 11[b]
(C) Time elapsed after cement mixing before cement injection	1.5 min	3.0 min
(D) Injection speed	0.4 mm/s (=0.226 ml/s)	0.8 mm/s (=0.452 ml/s)
(E) Average pore diameter of the filter	1.010 mm	1.905 mm
(F) Diameter of the side-hole	2 mm	4 mm

[a] Inner diameter: 3.7 mm; Product No. 180970, Somatex (Germany).
[b] Inner diameter: 2.3 mm; Product No. DBMNJ1104 T, Somatex (Germany).

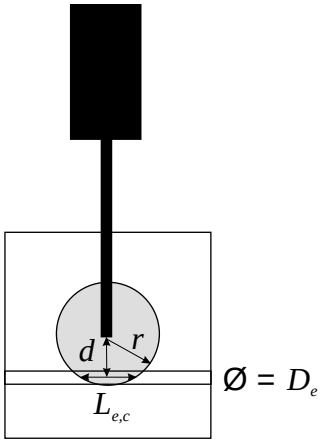

Fig. 4. Experimental model used to determine the effects of various parameters on the extravasation tendency. The cement is injected with a syringe and a needle into the center of the ceramic filter (square), at a distance d from the center of a cylindrical hole of diameter D_e. The sphere filled with cement has a diameter r, filling the cylindrical hole along a length $L_{e,c}$. In the experimental model, $d = 10$ mm. $D_e = 2$ or 4 mm. The ceramic filter is cubic and has a 5-cm long side.

octadecane (No. 74710, Fluka, Switzerland). Unfortunately, this compound was either hard below the melting point, or had a low viscosity above the melting point. As we wanted to have a viscous liquid roughly mimicking bone marrow, we replaced octadecane by plain cow butter. To fill the ceramic filters, the butter was melted at 60°C and poured over the ceramic filters. Then, the filters were impregnated under vacuum and left at room temperature for one day. All experiments were performed at room temperature. The ceramic filters were open-porous alumina blocks used for aluminium production (Filter plates Sivex; Pyrotek SA, Sierre, Switzerland). Two different pore sizes were purchased: (i) 900–1120 and (ii) 1640–2170 μm. The total porosity was 90%. The compressive strength of the filters was 1.1 and 1.5 MPa, respectively. The distribution of the cement in the ceramic filter was assessed radiographically.

4. Results

Two types of injection curves were observed (Fig. 5). First, the force was large enough to inject the cement into the filter (Fig. 5a). The curve was characterized by three parts: (i) a steep increase of force (up to 5 mm in Fig. 5a); (ii) an intermediate domain where the force increases moderately (from 5 to 23 mm in Fig. 5a); (iii) a final steep increase (after 23 mm in Fig. 5a). In the second type of injection, the force was not large enough to inject the cement into the filter (Fig. 5b). As a result, the force kept increasing without cement coming out of the needle. At a force of about 1200 N, the cement extravasated into the back of the syringe, finding a path between the plunger and the syringe walls. The experiment was then stopped. In most cases, a local maximum was seen along the curve (at 500 N in Fig. 5b—see also in discussion).

During the experiments, it was observed that the needles were deformed a lot, especially at high load. To determine this deformation, the syringes were filled with water, hermetically closed and submitted to a load. The measured displacements were large, reaching about 8–9 mm at 1000 N (Fig. 6). Smaller displacements were observed at a higher loading rate. Interestingly, a local maximum was also observed along the curves (at 900–1000 N in Fig. 6; see discussion).

In all samples where the cement was injected after 3 min through a gauge 11 needle, no cement came out into the ceramic filter. The applied force was obviously too small. Interestingly, when the gauge 11 needle was replaced by a gauge 7 needle, cement could be injected in all but one cases. Therefore, when the flow rate was kept constant, larger forces were required with thinner needles.

When extravasation occurred, the cement did not fill the ceramic filter, but flew out of the block, either along the needle or through the side holes (Fig. 7). To characterize the extravasation, the width of the radio-opaque zone at the injection point was compared to the width of the radio-opaque zone in the side-hole (see Appendix, Figs. 4 and 7). In the absence of extravasation, the width of the radio-opaque zone in the side hole was much smaller than the width of the radio-opaque zone at the injection point. When extravasation occurred, the opposite was observed. As no cement could be injected 3 min after the start of cement mixing with gauge 11 needle, only the results obtained with an injection time of 1.5 min were analyzed.

The statistical analysis of these results showed that the risk of extravasation increased significantly when the filter was filled with butter ($p < 0.0001$) or when the side-hole diameter was large ($p < 0.0001$). There was also a positive interaction between the latter two factors ($p < 0.001$). In other words, the effect of factor A (filter filling with butter) was amplified by a high level of factor F (i.e. size of the side-hole equal to 4 mm).

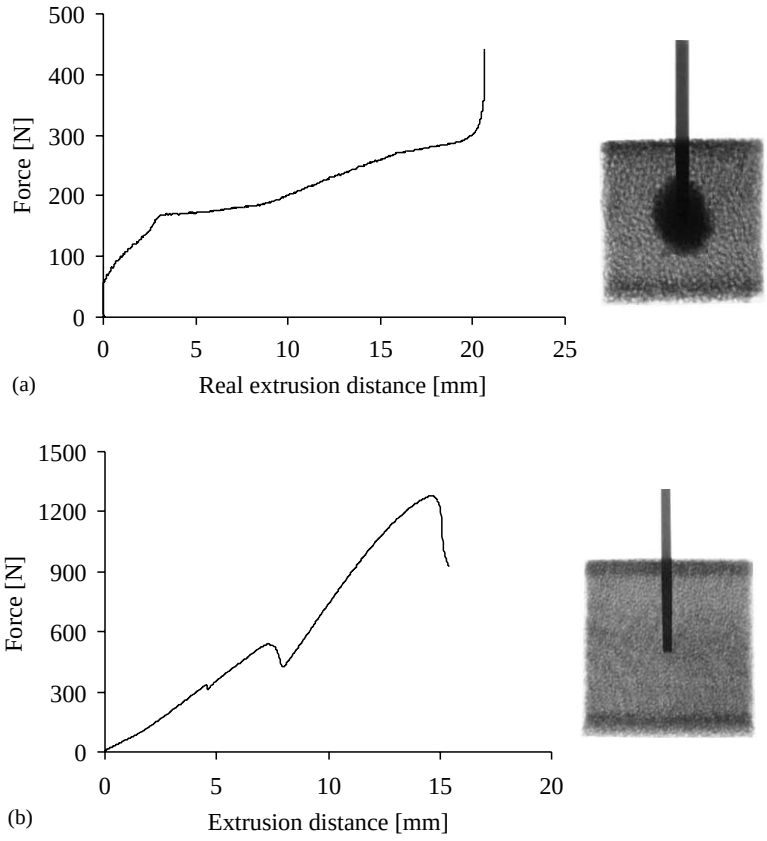

Fig. 5. Typical injection curves. Two behaviors are observed: (a) The cement is injected into the filter until no more cement is left into the syringe. The latter point is characterized by a steep increase of the force at a large extrusion distance; (b) The injection force is too small to inject the cement. The force only increases until the syringe breaks (at 1200 N). The curve is characterized by a small local maximum (here at 500 N). The end-result for each injection curve is represented radiographically, on the right-hand side of the injection curves.

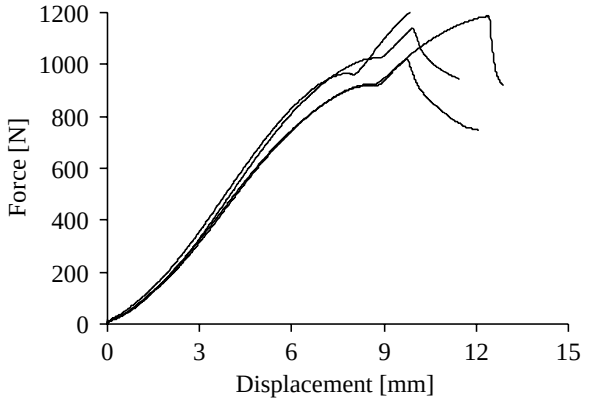

Fig. 6. Deformation curve of the syringe when filled with water and in the absence of flow (front part closed). There is always a local maximum on the curve, here at 900–1000 N. Top two curves: 0.8 mm/s. Bottom two curves: 0.4 mm/s.

5. Discussion

The large deformation of the syringes under loading prevented the systematic determination of the forces required for cement injection. An attempt was made to substract the deformation curve of the syringe to the measured injection curves. Unfortunately, the local maximum that was observed along the injection curves (e.g. at 500 N in Fig. 5b) was not always at the same position, which rendered the substraction useless. This local maximum was in fact due to the sudden flattening of the conical front part of the syringe. Obviously, new experiments should be performed with stiffer and stronger needles.

As the local maximum was never seen at a low force, the initial slope of the injection curves was measured after substracting the control curves. The analysis of these results showed that the initial slope was significantly increased by an increase of the time elapsed between the start of mixing and the time of injection ($p<0.001$). This phenomenon is obviously due to an increase of the cement viscosity (Eq. (1)).

The comparison of the different results shows that there is a good agreement between theoretical and experimental results. For example, the model predicted that the risk of extravasation would increase with a increase of the viscosity of the marrow (Fig. 3) and an increase of the diameter of the extravasation path, as observed experimentally. Therefore, the model can be used to improve our understanding of bone

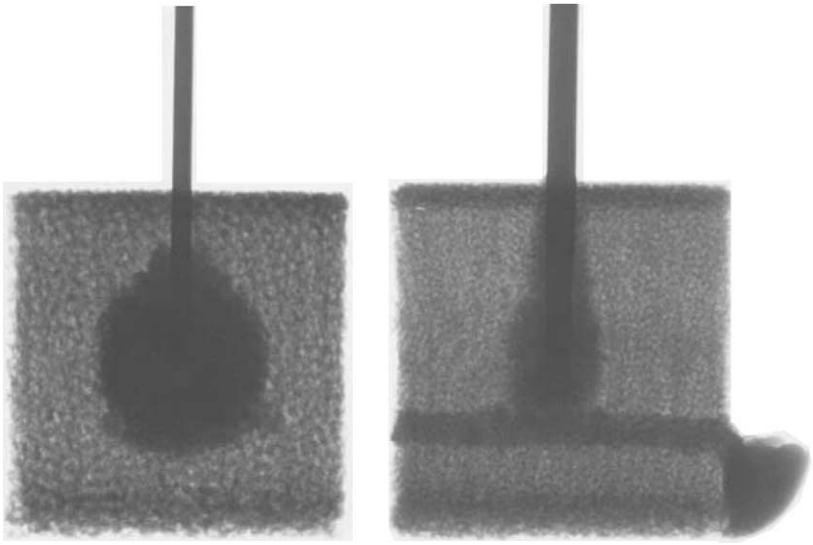

Fig. 7. Two cement filling patterns. The needle comes down from the top of the block. The dark area in the block is the area filled with PMMA cement. Left: perfect augmentation; Right: extravasation.

Table 2
Tendency of several parameters to produce cement leakage based on the theoretical model

Increase of	Tendency of leakage	
Bone permeability	Decreases	–
Cement viscosity	Decreases	–
Marrow viscosity	Increases	+
Diameter of extravasation path	Increases	+
Matrix porosity	Decreases	–
Size of the cavity at the injection point	Decreases	–
Pore size	Decreases	–

augmentation and cement extravasation, and in particular, to find new strategies to decrease bone extravasation.

The model predicted that several parameters had an influence on bone extravasation (Table 2): (i) the bone permeability; (ii) the cement viscosity; (iii) the marrow viscosity; (iv) the diameter of the extravasation path; (v) the bone porosity; (vi) the size of the injection cavity; and (vii) the bone pore size. Among all these factors, only one factor does not depend on the bone structure: the cement viscosity. Therefore, the easiest way to decrease the risk of cement extravasation is to increase the cement viscosity.

There are two simple solutions to increase the cement viscosity: (i) increase the time elapsed after the start of mixing or (ii) increase the powder-to-liquid ratio. In the former possibility, the time period during which the cement can be injected might be too short. In the latter possibility, a too high powder-to-liquid ratio might lead to inhomogeneities. In any case, the injection of a highly viscous cement paste requires the use of a large injection pressure.

High injection pressures can be obtained by reducing the syringe diameter: the pressure applied on the cement is indeed inversely proportional to the cross-section of the syringe. This approach is however limited by geometric aspects and by the extent of the force that a human hand can apply. Some experiments made with colleagues have shown that a force of 150 N seems to be a limit above which most people start shaking their hand. Therefore, when the force that has to be applied is larger than 150 N, an injection gun should be used.

Most people would rather use a bare hand to apply a force than an injection gun, because the link between the pressure applied by the hand and the resulting force applied on the cement paste is shorter. In reality, this is due to the fact that there are friction forces and deformations occurring on one hand in the cement paste and on the other hand in the whole injection system (e.g. plunger, side-walls of the syringe). These deformations and friction forces cannot be removed instantaneously by releasing the gun handle, and as a result, cement might flow even though no more pressure is applied on the gun handle. To decrease this effect, rigid injection systems should be used. Additionally, friction forces—for example, between the plunger and the syringe walls—should be minimized. Moreover, a small injection rate should be used as recommended by Barr and Barr [14]. Finally, the presence of air bubbles in the cement paste should be banished.

An easy way to restrict the injection flow rate is to use narrow syringes. The flow rate is indeed proportional to the inner cross-section of the syringe. An additional advantage of narrow syringes is that the pressure applied on the cement is inversely proportional to the cross-section of the syringe. Thus, narrow syringes enable the use of relatively low forces, which has a

positive effect on the control of the injection flow and hence on the extravasation occurrence.

The use of a high cement viscosity could have a deleterious effect: instead of penetrating into the bone structure, the cement could break it by "pushing" it away. In that case, there would not be any inter-digitation of the cement in bone, and possibly an absence of adequate mechanical support of the vertebral body. This aspect of the use of highly viscous cement pastes should be addressed in a future study.

Interestingly, Eq. (22) predicts that the risk of extravasation is not a function of the flow rate, Q_c. This result is true as long as the cement behaves like a Newtonian liquid. However, previous studies [11–13] have shown that PMMA cements are shear-thinning, i.e. the cement viscosity decreases with an increase of shear rate. As a decrease of the cement viscosity increases the risk of cement extravasation, it is recommended to inject the cement at a slow injection rate. Consequently, the cement velocity and the shear rate are low. This recommendation is particularly important at the start of the injection, when the cement velocity at the marrow-cement interface is large (Eq. (4)).

The predictions of our model are in agreement with surgical observations. Several authors [2,3,14,15] observed that the risk of extravasation decreased with an increase of the cement viscosity. Cotten et al. [2] observed a higher incidence of cement extravasation when bone augmentation was used to treat osteolytic metastases and myeloma, where the bone defects and hence the extravasation paths were large. More recently, Breusch et al. [17] compared the augmentation behavior of a low viscosity and a high viscosity cement in a sheep study. The latter authors observed that "the low viscosity cement yielded lower rates of cement penetration despite adequate and sustained pressurization. Cement applied at low viscosity state seems to take the path of least resistance into the venous system before more deeper cement penetration can occur. The use of high viscosity cement ran a higher risk of fat embolism but improved cement inter-digitation." These results are in accordance with the prediction of our model. In particular, the risk of extravasation increased with an increase of the injected volume and a decrease of the cement viscosity (Fig. 3).

To conclude, the results presented in this study are primarily related to PMMA. However, the model does not assume a priori that the cement consists of PMMA. Therefore, the same equations can be used for other types of cement, such as calcium phosphate cements.

6. Conclusion

In conclusion, a good agreement was seen between our analytical model, experimental results, and surgical observations. Therefore, our theoretical approach resulted in a very good model, able to predict qualitatively which factors are important for cement extravasation. Among these factors, the cement viscosity is certainly the most important, because it is the easiest to modify. The model predicts that cement extravasation can be drastically reduced by increasing the cement viscosity.

Appendix A

A.1. Permeability

The permeability of a porous matrix is intrinsic to the matrix and must be measured. However, some simple calculations can be used to understand how the permeability should vary as a function of porosity or pore size. If the porous matrix is made of cylindrical and parallel pores (Fig. 8), and if the flow of the fluid through the pores is controlled by the law of Hagen–Poiseuille, the flow rate in each of these pores is linked to the pressure drop in each of the pores, ΔP_p, by (Eq. (1)):

$$q_p = -\frac{\pi \Delta P_p D_p^4}{128 \mu_f L_p}, \tag{A.1}$$

where D_p is the pore diameter and L_p is the pore length (Fig. 8). The number of pores, n_p, is determined by the porosity of the matrix, p, according to:

$$n_p = \frac{4 p S_m}{\pi D_p^2}. \tag{A.2}$$

S_m is the surface area of the matrix (perpendicular to the pores). The global flow through the matrix is given by

$$Q_f = n_p q_p = -\frac{\Delta P_p D_p^2 S_m}{32 \mu_f L_p} p. \tag{A.3}$$

Eq. (2) can be re-written for the present case

$$\Delta P_p = -\frac{v \mu_f L_p}{K}. \tag{A.4}$$

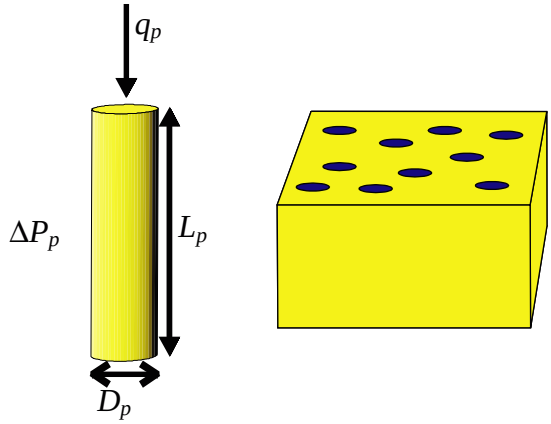

Fig. 8. Matrix constituted of cylindrical pores of radius R_p and length L_p. Flow through one cylindrical pore, q_p.

The average velocity of the cement through the matrix is

$$v = \frac{Q_f}{S_m}. \quad (A.5)$$

Therefore, assuming that the pore length is equal to the matrix length, the comparison of Eqs. (A.2)–(A.5) gives

$$K = \frac{pD_p^2}{32}. \quad (A.6)$$

In a normal case, the pores are not straight, so the pore length is longer than the thickness of the material. To take this into account, a "tortuosity" factor, τ, is introduced. The permeability becomes

$$K = \frac{pD_p^2}{32\tau}. \quad (A.7)$$

This equation shows that the permeability of the material increases with an increase of the porosity and the pore diameter. Interestingly, an homothetic increase of the size of a porous matrix by a factor of e.g. two should increase the permeability of this expanded matrix four-fold.

A.2. Extravasation length

In the experimental model, the length of the extravasation path filled with cement ($L_{e,c}$; Fig. 4) depends on the amount of cement injected into the ceramic filter. Assuming that the spreading of the cement is spherical (as in Fig. 4), the length filled with cement is

$$L_{e,c} = 2\sqrt{r^2 - d^2}. \quad (A.8)$$

Obviously, this result is only valid if $r \geqslant d$. If r is smaller than the distance d, the length $L_{e,c}$ is equal to zero, i.e. there is no cement in the extravasation path.

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