

Investment casting for oral cavity applications: peculiarities and problems

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The aim of the investment casting in dentistry is the attainment of a custom-made medical device. As a consequence, the medical device must satisfy requirements that are critical for patient's health, such as material bio-compatibility and device effectiveness. In fact, the medical devices should recreate the original oral environment and – at the same time, should not show health-threatening side effects, both at short and long times, with long times taken as ten years at least [1]. As a consequence, the investment casting must be applied to appropriate materials and the actual application environment of the device should be taken into account. The device must have form and dimensions able to support the chewing loads, to match the patient oral cavity and to stay there for a long time. The work is aimed at summarizing the properties of the alloys and the investment casting processes used in dentistry.

Keywords: precious metals, biomaterials, defects, fracture, solidification, foundry, casting, failure analysis, metallography, scanning electron microscopy, material selection, properties

MATERIALS

Dental alloys

The first application of the investment casting in dentistry occurred in 1891 by Martin, who realized gold castings in a lost-wax mould. Later on, new materials were developed until the present-day precious metal alloys for resin-base aesthetical coatings. Furthermore, precious metal alloys for ceramic aesthetic coatings were developed since 1950, by Weinstein. Today, several hundreds alloys belonging to these two classes are on the market as well as Ni-Cr alloys, Cr-Co alloys, Ti and Ti alloys. The great number of available materials makes it difficult to summarize their physical and mechanical properties [2], [3], [4], [5]. The lack of data regarding other properties like fatigue resistance, or microstructure increases the difficulty. The wide variety of available alloys implies a complex variety of microstructures as much, to which correspond only a few studies on phases and structural constituents.

Investment casting dental alloys can be divided into three main classes: crown and bridges alloys, for resin aesthetical coatings; alloys for metal-ceramic used for ceramic aesthetic coatings and multipurpose alloys, for resin or ceramic with low sintering temperature. This classification relates to the final application of the metallic substructure obtained by investment casting. The substructure can be coated by resin or ceramics to restore the original tooth appearance. A fourth kind of alloys exists, used for removable prostheses obtained by investment casting.

The following tables report the alloy general properties. They are not exhaustive, but give a general overview on chemical composition and mechanical properties. The main requirements that dental alloys must satisfy can be found in the following standards: ISO 9693:1999 (Metal-ceramic dental restorative systems), EN ISO 1562:1995 (Dental casting gold alloys), EN ISO 8891:2000 (Dental casting alloys with noble metal content of at least 25% but less than 75%), ISO 6871:1994 (Dental base metal casting alloys. Part 1: Cobalt based alloys. Part 2: Nickel based alloys).

Clinically important properties of dental alloys

Usually, the choice of the alloy by the dentist or by the technician is made on a low-cost basis. On the contrary, it should be made on the basis of the best chemical, physical and biological characteristics related to the specific clinical condition. Unfortunately, the necessary complete information is rarely available, hence the choice is made on a purely economic basis.

Alloy microstructure

One of the most important property - often unavailable, is the knowledge of the alloy microstructure. The grain dimension and the distribution of second phases and structural constituents are among the major microstructural features. The grain size, determined by the alloy investment casting, can be controlled by the grain refiners [6]. Grain refiners are high fusing elements, like Ir, Ru, Re and Rh and are used in concentrations from about ten p.p.m. to 1 %. Since they remain solid during the alloy casting process, they act as crystallization nuclei during the alloy solidification, giving rise to sufficiently small grains. Grain dimensions vary from 10 to 90 μm in precious metals alloys with a correct refinement. In this case grains are almost equiaxial in shape, without elongation in preferential directions. Larger and dendritic grains can be found in base metal alloys. In these cases the grain size can be even of one millimetre or more, equal to the whole section of the device.

Grain size is crucial because it affects the clinical properties of the medical device. In fact, mechanical properties are improved by reducing the grain size. Smaller grains correspond to a higher ultimate elongation and a higher ultimate tensile strength. In the clinical perspective, this allows the use of a more reliable material when chewing loads are applied. Additionally, small grains improve the alloy chemical homogeneity and the corrosion resistance, as a result. Then, an alloy with a well-refined grain gives the best results in investment casting. In fact, shrinkage porosity diminishes in this case. As a result, the mechanical properties of the device are improved, as well as the corrosion resistance, since the surface porosity is lowered. In fact, corrosion resistance can be threatened by surface micro-cavities that act as possible corrosion initiation sites by differential aeration or as bacterial plaque stagnation sites [7], [8], [9], [10], [11].

It is also noteworthy that the grain size obtainable by investment casting is not a-priori guaranteed by the alloy quality. It depends on the casting methods, on the presence of alloy

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	Composition (mass %) X = < 1%													ΔT (°C) (5)	HV5/30
	LEGA	Au	Pt	Pd	Ag	Cu	In	Sn	Ga	Zn	Ir	Fe	Other		
Yellow Ceramic Gold Alloys	1	87	10				X				X	X	Rh	1045÷1140	150
	2	86	10.8				X			1.5			Mn,Ta,Rh	1050÷1140	190
	3	81.5	13.4	X	X	X				X			Ir	1070÷1120	170
	4	81	17.5				X				X		Ta	1090÷1260	145
	5	77	19		1.7					1.7	X		Ru	1060÷1180	180
	6	75		13	10		X	X			X			1175÷1230	140
	7	74	4	7.8	11.2		2	1			X			1075÷1160	195
White Ceramic Gold Alloys	8	65.8		25.7			8.1		X		X			1135÷1225	220
	9	62		24.5	7		6				X		Ru, B	1175÷1260	220
	10	52		29.8	14.2		3	X					Ru, B	1200÷1260	210
	11	51.5		38.4			8.5		X				Ru	1130÷1290	245
	12	50.7	1	38.8			7.4		2				Ru	1230÷1310	240
	13	50.5		35.3	5.5		6	X		X			Ru	1215÷1300	190
	14	50.6	X	26.6	17.5		2	X			X	X	Mn, Ru	1175÷1260	190
	15	45.4		40	5.4	X	8	X	1.5				Ru	1210÷1270	240
	16	45		44.8			5.2	3	X			X	B	1200÷1290	250
White Ceramic Precious Metals Alloys	17	43.1		30.7	19.8	1.5		5.9				1.5	B, Re	1140÷1230	220
	18	40		44.8	5		6.6		1.5	1.9			Ru, Li	1150÷1250	230
	19	32.1	3	42	13.9		4	2		2			Ru, B	1210÷1295	190
	20	29.1	13.5	23.2	27	X	4			X	X	X	Ru, Li	1140÷1210	250
	21	13.2		57.8	17		X	3.5	X				Re, B	1155÷1275	220
White Ceramic Palladium Alloys	22	2		79		10			9				Ru	1090÷1215	320
	23			78.9		10	4		7	X			Ru, B	1195÷1210	330
	24	6		74.8	6.5		6.5		5.9				Ru, Li	1100÷1300	270
	25	2		74.7		9.4	6.5		4.9	2.2			Ru	1145÷1230	315
	26	1.9		72.1		14.5	4	4		3.5			Ru, B	1200÷1260	250
	27	2.1		67	16.9	5.8		3.7	4.2				Ru	1130÷1230	270
White Ceramic Palladium-Silver Alloys	28			63.4	26.1		X	7	X				Ru	1140÷1280	220
	29			61.3	24.5		2.1	10		X			Ru, B	1160÷1250	310
	30	0,3		59,9	27,6		4	7	1				Ru	1170÷1270	300
	31			58	30		3	4	4				Ru, B	1130÷1265	290
	32			56.1	33.5			7.8	2.5				Ru, Li	1065÷1200	280
	33			53.3	37.4		X	8.6					Ru	1190÷1250	240
Multipurpose Alloys (9)	34	75	9		12.4					2.1	X		Rh, Ta	1000÷1080	190
	35	73	1.7	5.7	16			X		3	X			980÷1070	130
	36	72	9.3		10	4.8	3.1	X		X	X			900÷990	170
	37	55.2	11.4	3	23		3.5			3	X		Rh, Ta	940÷1080	170
	38	55		10	28.7		4.2	X		1.5	X			970÷1050	200
	39			40	51.9		2	2		4			Ru	1070÷1140	160
(1) All data as cast (tutti i dati nelle condizioni di materiale dopo fusione) (2) 0,2 % offset (3) C.T.E. : Coefficient of linear Thermal Expansion from 25 °C to 500°C (X 10-6 / °K). (Coefficiente di dilatazione termica lineare medio tra 25 e 500 °C (X 10-6 / °K)) (4) U.T.S. : Ultimate Tensile Strength (5) Melting Range (6) Specific Gravity (7) Elongation %. (Allungamento percentuale a rottura) (8) Modulus of Elasticity (9) For resins and high expansion low sintering temperature ceramics															

Table I – Properties of some alloys for metal-ceramic restorations.

Yield Strength (MPa) (2)	U.T.S. (MPa) (4)	ΔL (%) (7)	ρ g/cm ³ (6)	CTE (3)	E (MPa) (8)
370	450	10	18.5	14.5	90000
450	550	6	18.7	14.2	95000
315	435	18	19.1	14.4	95000
335	495	14	18.5	14.2	98000
410	500	6	18.7	13.8	101000
430	600	7	16	14.9	95000
430	570	9	17	14.8	88000
500	550	10	15.2	14.2	120000
450	675	5	14.7	14.6	107000
550	690	10	13.8	14.2	118000
570	720	13	13.9	13.7	120000
570	780	20	14.5	13.7	124000
500	590	7	14.2	14	120000
500	700	15	14.1	14.7	119000
550	690	7	13.8	13.9	114000
550	690	10	13.5	13.6	117000
535	700	9	13.9	14.6	120000
530	740	18	13.5	14.2	124000
395	600	14	13.9	14.5	114000
600	740	12	13.8	15	138000
690	970	16	12.2	14.2	130000
910	1130	16	10.6	13.8	100000
690	895	18	10.5	14.2	97000
590	775	25	11	13.9	111000
760	920	10	11.1	14	105000
675	1100	20	10.6	13.8	98000
800	950	9	11.3	14.1	125000
500	780	38	12.6	14.8	131000
600	850	15	11.3	14.5	130000
700	890	7	11.3	14.5	134000
715	905	9	10.9	14.6	129000
710	920	16	11.3	15	125000
530	660	7	11.3	14.7	137000
540	700	8	16.7	15.8	85000
660	720	5	15.7	16.3	98000
340	530	13	16.6	16.6	99000
390	550	6	14.7	16.3	100000
500	590	4	14.2	16.9	115000
285	550	10	10.6	16.6	100000

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scraps coming from previous processes and on the geometrical shape of the wax model. Particularly, if more than 50 weight % of alloy scraps are added, grain size degrades as a consequence. Some of the grain refiners are lost, due to their tendency to adhere to the crucible walls during melting or to agglomerate during alloy re-casting, losing their refining properties. Additionally, grain size depends on the device thickness. Lower size corresponds to smaller thickness, higher size to higher thickness.

The distribution of phases and constituents is critical from a clinical point of view. A single phase material has a homogeneous chemical composition, while a multi phase alloy has a chemical composition changing on a microstructural scale. In this case the corrosion resistance may decrease, especially when one or more elements with a low chemical stability in the oral cavity, concentrate in some phases. In fact, the initiation of galvanic corrosion between different phases occurs frequently, especially if their electrochemical nobility is different. Mechanical properties can be influenced by non homogeneous microstructures as well. For example, if a phase or a structural constituent is brittle, both mechanical properties and fatigue resistance decrease. The non homogeneous microstructure of the alloy is influenced by the investment casting methods also. The temperature of casting and of the dental casting mold as well as the cooling times can determine the volume fraction of some structural constituents. At last, an appropriate thermal treatment may homogenize the alloy, by reducing the volume fraction of certain structural constituents or even leading to a single phase alloy.

Mechanical properties

Mechanical properties are usually available for all commercial alloys. Some of them are significant for a good clinical performance of the dental alloy. As an example, the elastic limit must not be exceeded, in order to avoid a permanent distortion of the medical device. The typical location for this failure is between pontics in a multiple-unit fixed partial denture. Alloys with an elastic limit higher than 300 MPa resist against permanent deformation in the oral cavity in the majority of clinical cases.

Alloy hardness is not of primary concern from a clinical point of view. However, it should be remembered that if the alloys is exposed to the oral environment, it can be scratched if its Vickers hardness is less than 130 HV, or can scratch the antagonist teeth if its Vickers hardness is more than 340 HV. The elastic modulus of dental alloys should be as high as possible in order to avoid bending, that could cause the detachment or the ceramic cracking in metal-ceramic alloys. Resin alloys are susceptible to undergo this drawback also. Detachment is more likely to occur than coating cracking in this case. Usually, the elastic moduli of precious metal dental alloys range from 90 to 130 GPa. Elastic moduli values are higher for high palladium content alloys and lower for high gold content alloys. These values are acceptable in the majority of clinical cases. However, in the case of long-span metal-ceramic restoration, it may be necessary to use Ni-Cr or Co-base alloys. Elastic moduli range from 150 and 230 GPa in these cases.

Alloy colour

The alloy colour is not important from a clinical point of view. However it played a critical role historically, influencing the choice of the alloy. Yellow was always associated with a high gold concentration, considered as a high quality proof. However, the clinical performance depends on other parameters, as already discussed. Furthermore, yellow alloys with very low or zero gold content are available nowadays (see alloy 52 in the previous tables). The corrosion resistance is low for these alloys, but good even in white alloys containing palladium and gold.

	Composition (mass %) X = < 1%													ΔT (°C) (5)	HV5/30
	LEGA	Au	Pt	Pd	Ag	Cu	In	Sn	Ga	Zn	Ir	Fe	Other		
Yellow Crown & Bridge Casting Alloys	40	90	8.9				X				X			1070÷1165	100
	41	77.4	1.3		12.2	8.5	X			X	X			900÷940	125
	42	75	9		12.5					X	X		Rh (1%),Ta	1005÷1075	130
	43	74.5		3.5	11	10.4				X	X			930÷960	120
	44	70	5.6		13.5	9.9				X	X			910÷955	240
	45	62.9		2.9	22.9	9	1.3			1	X			890÷955	130
	46	60		5	22.5	10.6	X	X		X			Ge	865÷920	260
	47	58.7	1.7	3.3	22.3	12.7				1.2	X			870÷910	280
	48	53		6	24.5	15				X	X		Ge	845÷890	290
	49	50		5	32.5	11.5				1	X			845÷925	160
	50	46.1		6	39.5	7.4				1	X			880÷965	220
	51	40		4.9	45.8	8.3				1	X			855÷925	125
	52	20		20	39.9		18			2	X			880÷925	125
White Crown & Bridge Precious Metals Alloys	53	2		34.9	30		30			3	X		Ca, B	1040÷1120	180
	54	30	1	9.9	38.1	19.9				1	X			845÷885	180
	55	15	1	18	55	9.9				X	X			935÷1010	165
	56	11	0.5	19.9	54.5	12.5	X			1.5				910÷990	190
	57	2		25	68		3			2	X			1010÷1095	150
	58			20	75		3			1,95	X			925÷980	135
	59			25	58.3	14.6	2						Ru, Li	910÷990	190

(1) All data as cast (tutti i dati nelle condizioni di materiale dopo fusione); (2) 0,2 % offset (3) C.T.E. : Coefficient of linear Thermal Expansion from 25 °C to 500°C ($X 10^{-6} / ^\circ K$). (Coefficiente di dilatazione termica lineare medio tra 25 e 500 °C ($X 10^{-6} / ^\circ K$)); (4) U.T.S. : Ultimate Tensile Strength; (5) Melting Range; (6) Specific Gravity; (7) Elongation %. (Allungamento percentuale a rottura); (8) Modulus of Elasticity

Table II – Properties of some alloys for crown and bridge restorations.

	Composition (mass %) X = < 1%													ΔT (°C) (5)	HV5/30
	LEGA	Ni	Cr	Mo	Be	Co	Al	Si	Fe	Nb	Ta	W	Other		
Non Precious Alloys for Ceramics and Removable Prosthetics (Base metal alloys)	60	78.2	13	5	1.8								X (2%)	1290÷1345	260
	61	76	14	6	1.8		2						X	1230÷1290	240
	62	72	15	9	1.8		2						X	1205÷1260	200
	63	68.5	15.5	14			1	X					Mn(X), X	1330÷1390	190
	64	54	22	9					4	4	4		X (3%)	1220÷1330	240
	65		31.5	5		59.5		2					Mn (1%), X	1240÷1350	280
	66		28	6		64							X (2%)	1300÷1370	380
	67	0.15	31.5	2.2		65.4							Mn,Fe,Al,C	1240÷1350	290
	68		25	7		61		1.5				5	Mn,Ce,N,C	1325÷1415	340

(1) All data as cast (tutti i dati nelle condizioni di materiale dopo fusione); (2) 0,2 % offset; (3) C.T.E. : Coefficient of linear Thermal Expansion from 25 °C to 500°C ($X 10^{-6} / ^\circ K$). (Coefficiente di dilatazione termica lineare medio tra 25 e 500 °C ($X 10^{-6} / ^\circ K$)); (4) U.T.S. : Ultimate Tensile Strength; (5) Melting Range; (6) Specific Gravity; (7) Elongation %. (Allungamento percentuale a rottura); (8) Modulus of Elasticity; (9) Annealed

Table III – Properties of some base metal alloys.

Titanium Alloys	Composition (mass %) X = trace elements								ΔT (°C) (5)	HV5/30
	LEGA	Ti	Al	V	Fe	O	H	N		
	69	100			X	X	X	X	1670	175
	70	90	6	4			X	X	1605÷1660	230

(1) All data as cast (tutti i dati nelle condizioni di materiale dopo fusione); (2) 0,2 % offset; (3) C.T.E. : Coefficient of linear Thermal Expansion from 25 °C to 500°C ($X 10^{-6} / ^\circ K$). (Coefficiente di dilatazione termica lineare medio tra 25 e 500 °C ($X 10^{-6} / ^\circ K$)); (4) U.T.S. : Ultimate Tensile Strength;

Table IV – General properties of titanium and titanium alloys used in dentistry.

Yield Strength (MPa) (2)	U. T. S. (MPa) (4)	ΔL (%) (7)	ρ g/cm ³ (6)	E (MPa) (8)
190	250	10	19.4	90000
290	450	45	15.9	90000
275	435	22	16.7	100000
210	430	40	15.5	75800
650	820	12	15.4	110000
275	400	20	14.2	75500
700	810	6	14.2	112000
900	990	3	14	115000
600	720	4	13.2	114000
340	510	30	13.4	74000
600	760	13	13.1	100000
240	410	20	12.4	81000
340	420	7	11.8	50000
300	340	4	10.5	51000
480	510	6	12.1	90000
320	460	14	11.3	115000
450	550	11	11	106000
300	430	18	10.5	82500
260	310	14	10.5	82000
385	490	20	10.5	100000

Yield Strength (MPa) (2)	U. T. S. (MPa) (4)	ΔL (%) (9)	ρ g/cm ³ (6)	E (MPa) (8)
515	920	13	7.8	185000
550	1135	15	7.8	192000
550	895	10	7.8	192000
300	670	35	8.5	160000
360	580	6	8.6	160000
450	765	9	8.8	160000
550	750	10	8.3	155000
445	765	9	8.8	165000
700	900	7	8.6	200000

Yield Strength (MPa) (2)	U. T. S. (MPa) (9)	ΔL (%) (7)	ρ g/cm ³ (6)	E (MPa) (8)
345	515	15	4.5	105000
890	950	14	4.4	120000

(5) Melting Range;
 (6) Specific Gravity; (7) Elongation %. (Allungamento percentuale a rottura)
 (8) Modulus of Elasticity
 (9) Annealed

Other properties

Investment casting is a critical phase in the production of a custom-made medical device.

It should be remembered that the realization of the device involves the combination of the investment casting product with other materials like resin or ceramics. In the latter case, the dental alloy should be compatible with the ceramic and with the ceramic sintering, performed during the application process. Sintering is a high temperature process - temperatures of about 980°C for conventional ceramics and 850°C for low fusing ceramics. To avoid detachment or breaking of the ceramic layer, the thermal expansion coefficient of the alloy should be slightly higher than that of the ceramic, in order to give the ceramic a low compression state. From a clinical point of view no stresses incompatible with the use in the oral cavity should exist in the ceramic parts. Particularly, no tensile stresses should be present on the ceramic surface, since they can enhance crack initiation and propagation. Usually, conventional dental ceramics have a thermal expansion coefficient between 13 and 14 x 10⁻⁶/°C.

At last, the alloy must contain elements that give rise to an oxide layer at the ceramic interface, to facilitate a chemical bonding with the ceramics.

As regards fatigue properties of dental alloys, only few data are available at present [12], [13], [14], [15]. However, the investment casting process may affect them seriously. For example, shrinkage or gas porosity and microstructural inhomogeneities may act as crack initiation sites, as previously discussed.

The melting temperature and the castability are important to the processor of the alloy. Castability cannot be deduced from the melting range alone. It also depends on the adopted casting technique. Usually the melting temperature is available, but a knowledge of castability for a given alloy is only a matter of experience.

Microstructure

Optical micrographies of different dental alloys are reported in the following as examples.

Micrographies relate to specimens obtained by investment casting, in the as cast condition. Specimens were prepared for metallographic examination to a 1 μm diamond paste polishing and chemically etched to reveal microstructure.

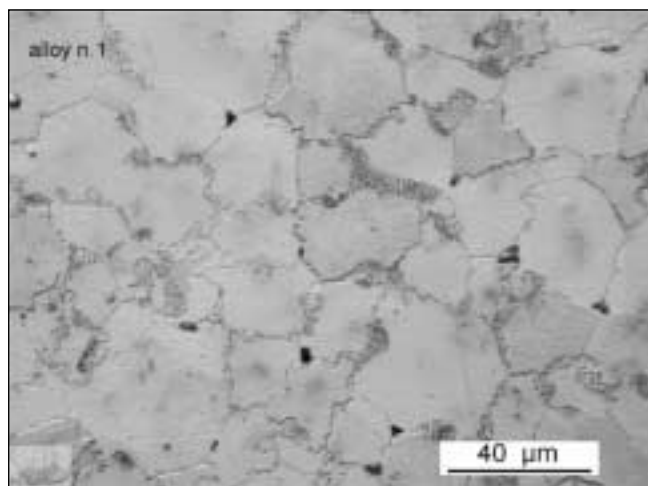
The images – obtained with a Zeiss Axiomat metallographic optical microscope – show the wide variety of microstructures of these materials. The elements concentrating in some phases or structural constituents are also indicated. The elements were identified by an EDAX energy dispersive X-ray spectrometer, known as X-ray microanalysis, attached to a PHILIPS ESEM XL30 TMP W Scanning Electron Microscope.

SPECIAL ASPECTS OF THE PRELIMINARY INVESTMENT CASTING PROCEDURES

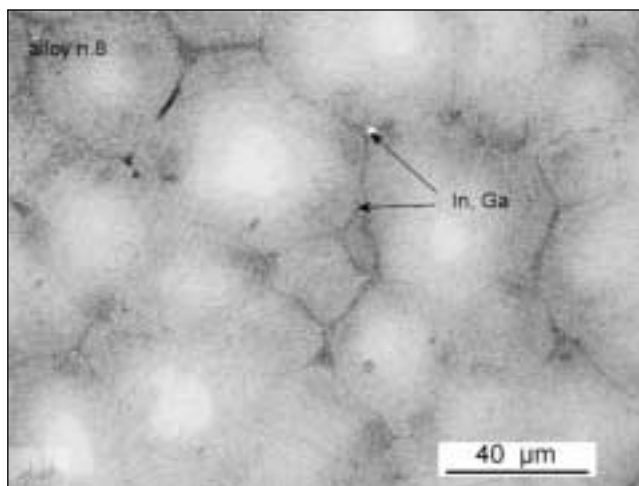
The first step in the investment casting process is the preparation of the wax pattern (fig. 1). The dimensional and morphological features of the pattern depend on the specific clinical situation, besides the kind of the chosen prosthesis (removable prosthetics, metal-ceramic bridges/crowns or crown and bridge alloy/resin prosthetics). A lot of aspects should be taken into account during wax modelling. Among them, the thickness of the different zones are relevant to the mechanical resistance of the device [16].

The most striking aspect of investment casting in dentistry is the accuracy of reproduction of the wax pattern by the metal structure. In fact, the accuracy by which the device fits the oral cavity is essential for its duration.

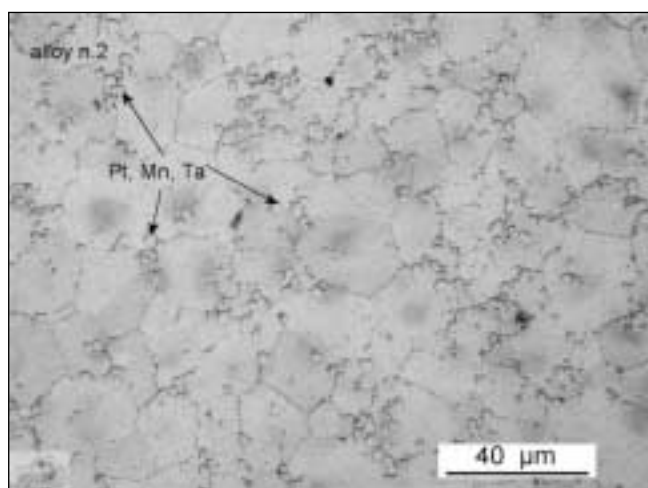
An optimal fit leads to less plaque accumulation at the mar-



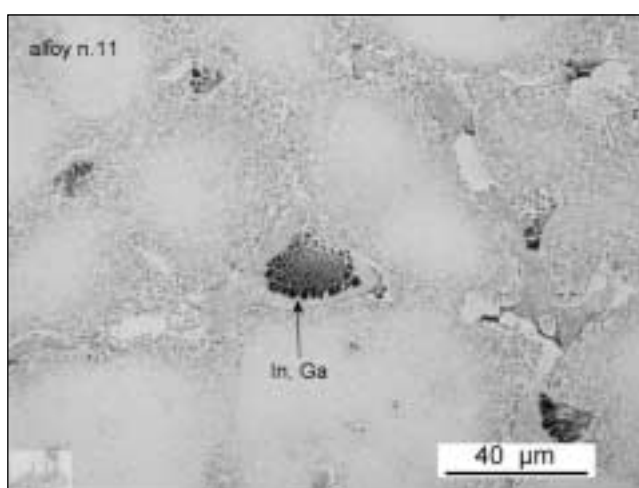
Alloy n.1: 87 Au, 10 Pt



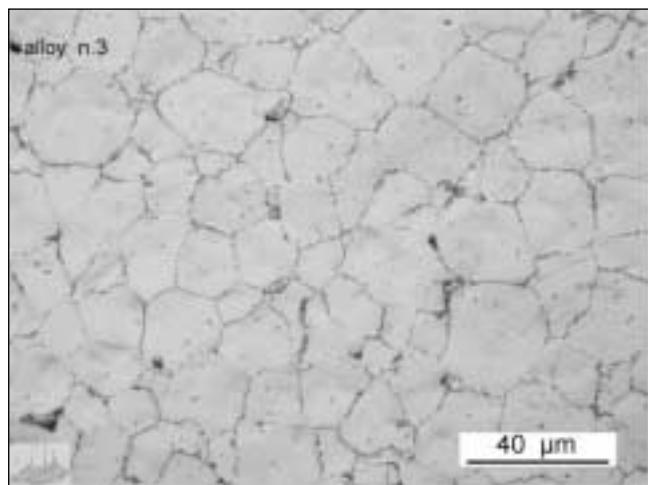
Alloy n.8: 75 Au, 13 Pd, 10 Ag



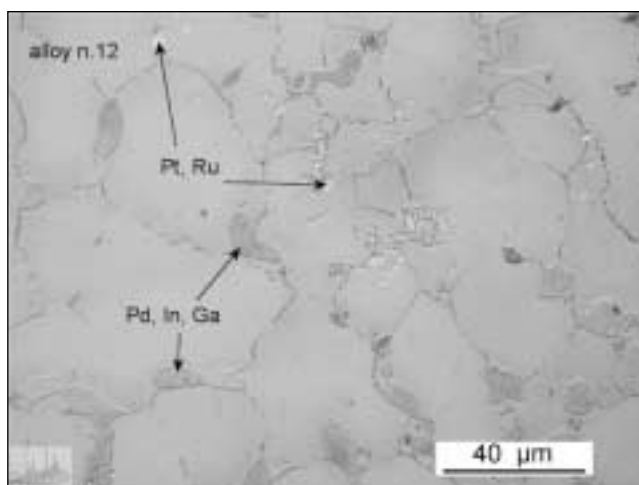
Alloy n.2: 86 Au, 10,8 Pt



Alloy n.11: 51.5 Au, 38.4 Pd, 8.5 In



Alloy n.3: 81.5 Au, 13.4 Pt



Alloy n.12: 50.7 Au, 38.8 Pd, 7.4 In

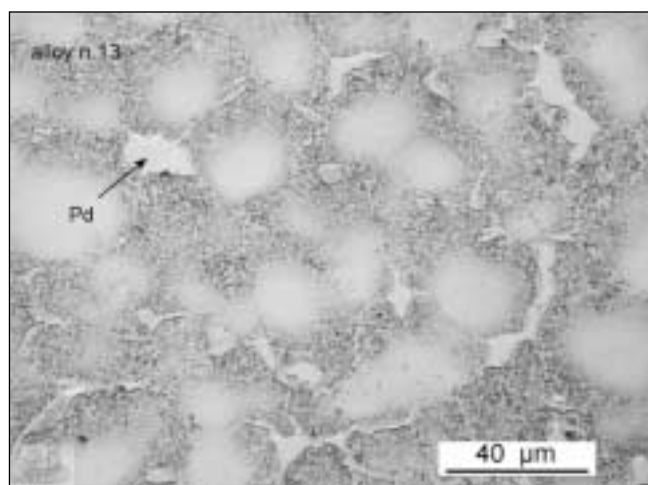
ginal area, better mechanical stability, less cement for prosthetic fixing and better aesthetical results.

Of course, it is necessary to quantitatively define this aspect. Unfortunately, literature does not provide a unanimous evaluation of the problem. In any case, figure 2 gives the “marginal gap” parameter, that is commonly used to measure the accuracy of fit.

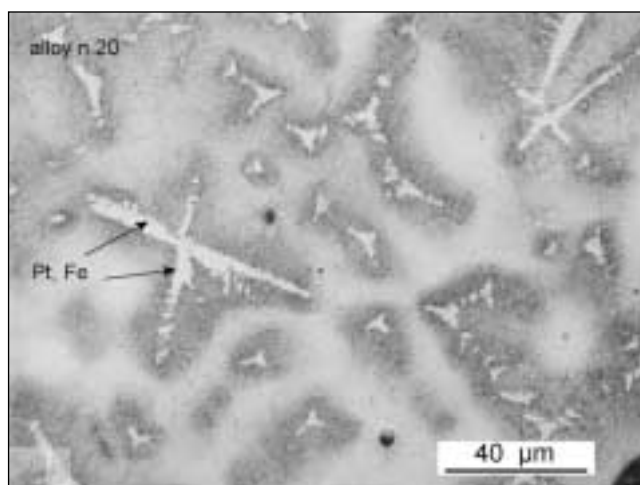
According to a clinical investigation [17] on 56 patients two weeks after the prosthetics insertion, the mean maximum marginal gap between restoration margin and tooth structure

was $64 \pm 18 \mu\text{m}$ for gold restorations. Literature does not give a commonly accepted value for the maximum biologically acceptable marginal gap. According to some authors [18], it should be lower than $50 \mu\text{m}$, while according to others [19] a marginal gap of $100 \mu\text{m}$ can be accepted. Anyway, it should be remembered that the marginal gap measurement in clinical cases may be hampered by the presence of the gingival tissue. Often, the marginal gap measurement is performed on replicas.

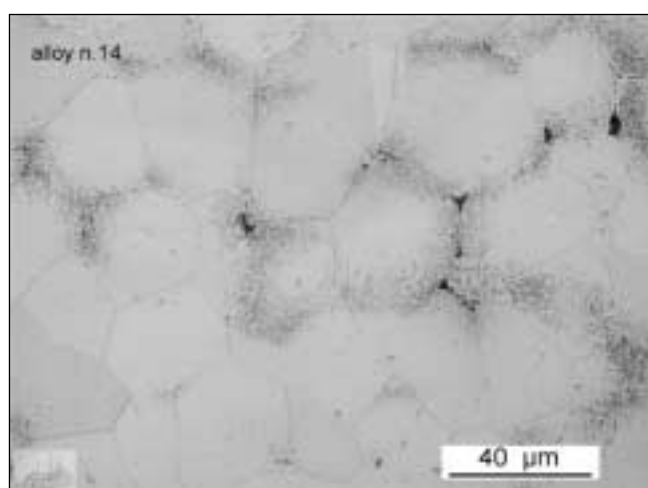
At any rate, the accuracy of fit is influenced by a lot of ope-



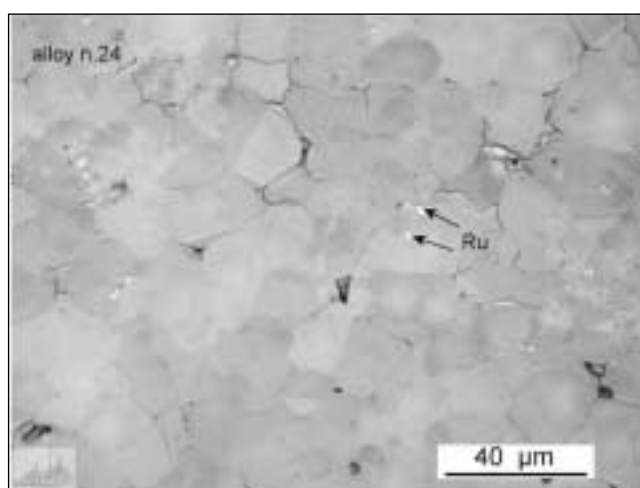
Alloy n.13: 50.5 Au, 35.3 Pd, 5.5 Ag, 6 In



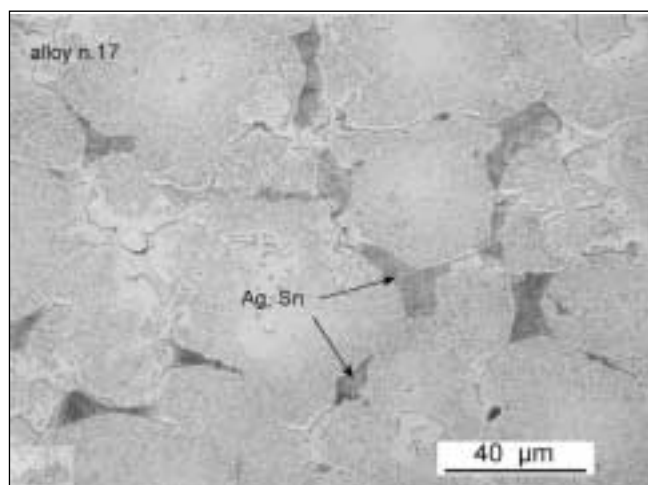
Alloy n.20: 29.1 Au, 13.5 Pt, 23.2 Pd, 27 Ag



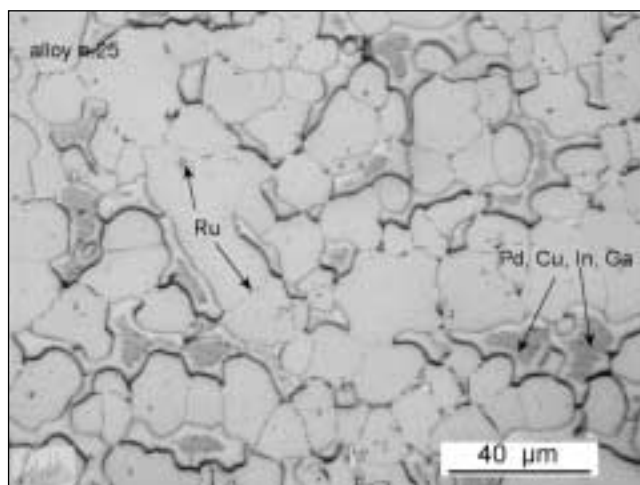
Alloy n.14: 50.6 Au, 26.6 Pd, 17.5 Ag



Alloy n.24: 74.8 Pd, 6.5 Ag, 6 Au, 5.9 Ga



Alloy n.17: 43.1 Au, 30.7 Pd, 19.8 Ag



Alloy n.25: 74.7 Pd, 9.4 Cu, 6.5 In, 4.9 Ga

rations, such as the quality of the preparation of tooth, the getting of the impression in the oral cavity, the quality and the kind of wax manufacture [20], [21] and, at last, the investment casting procedure, from investment to casting [22], [23], [24], [25], [26], [27], [28], [29], [30].

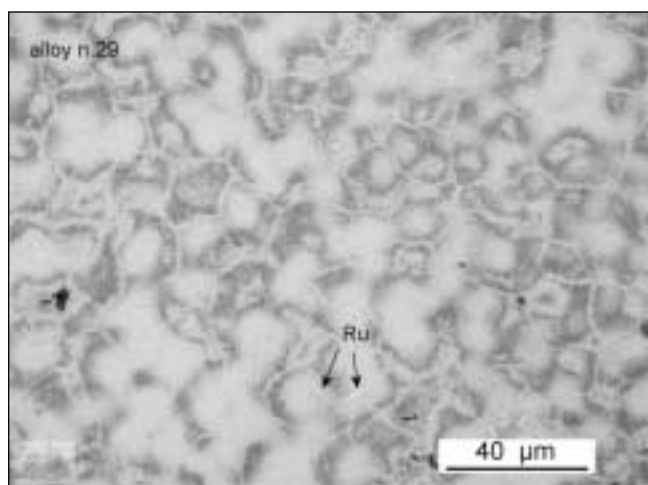
In this last step, a balance between the expansion and contraction of wax, investment and alloy, caused by the different operations, must be achieved.

Wax contracts during the solidification occurring in the wax pattern construction.

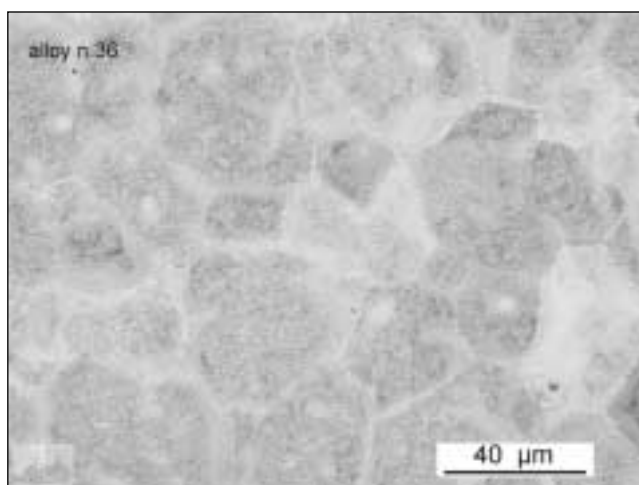
Later on, it expands during the investment setting and before burning, when the dental casting mold is put into the furnace and brought to the final pre-heating temperature – between 650 and 900°C, according to the alloy casting temperature.

On the other hand, the alloy contracts during solidification [16], [31], [32]. The investment expands during setting and the pre-heating of the dental casting mold [33], [34], [35], [36], [37], [38].

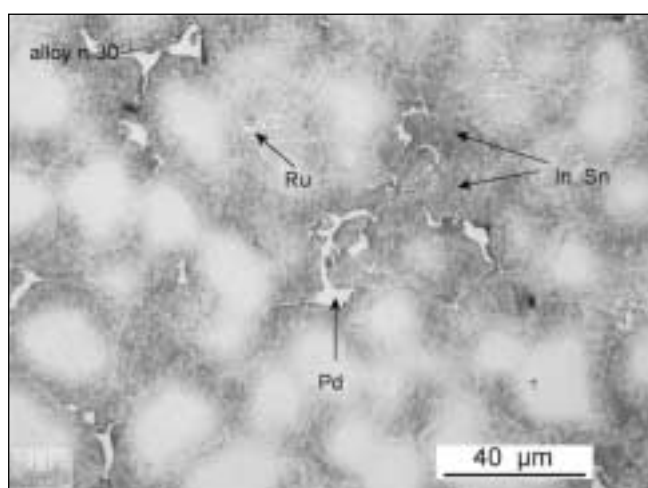
The most of the accuracy is obtained by counterbalancing



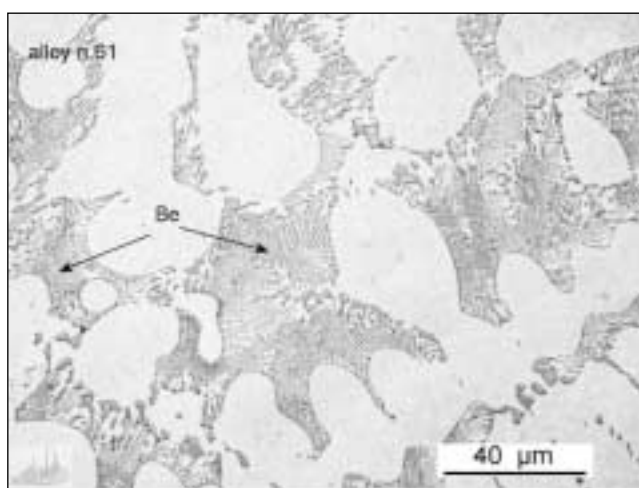
Alloy n.29: 61.3 Pd, 24.5 Ag, 10 Sn



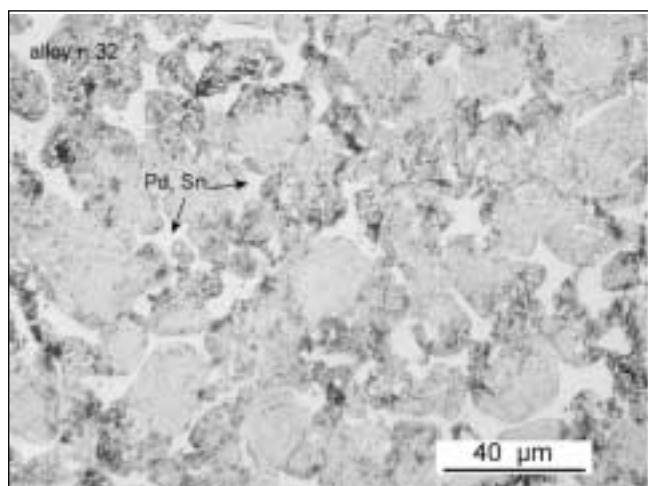
Alloy n.36: 72 Au, 9.3 Pt, 10 Ag



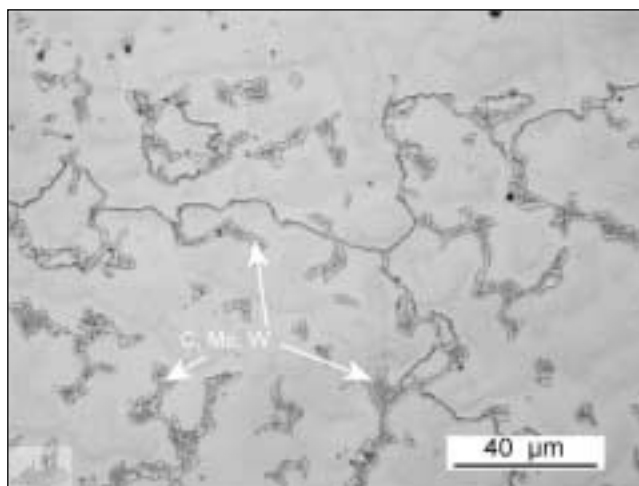
Alloy n.30: 59.9 Pd, 27.6 Ag, 7 Sn



Alloy n.61: 76 Ni, 14 Cr, 6 Mo, 1.8 Be



Alloy n.32: 56.1 Pd, 33.5 Ag, 7.8 Sn



Alloy n.68: 61 Co, 25 Cr, 7 Mo

the alloy solidification contraction with the investment expansion. The alloy contraction may range from 2.2 % to 1.7 %. Phosphate bonded commercial investments can counterbalance it, thanks to the expansion control achievable by properly dosing up the mixing liquids, made of aqueous colloidal silica suspensions. (fig. 3).

The investments used for precious metals alloys applied for ceramic aesthetical coating or base metal alloys are phosphate bonded. As regards precious metal alloys for resin aesthetical coating, gypsum-base investments are used.

MELTING AND CASTING TECHNIQUES

At present, the most widely applied casting technique is the centrifugal technique based on torch heating (fig. 5). Usually, oxygen and propane are applied, however other less safe gases are used also. In this case the alloy is melted into a refractory crucible and centrifugally pushed into the dental casting mold placed at the crucible exit hole.

The mold is pre-heated for 1 hour at a the right temperature, according to the alloy used. The mold is put in position few



Fig. 1 – Figures show some steps of the wax pattern preparation. On the left some recommended sprue dimensions are given.

Fig. 1 – Le figure mostrano alcune fasi della preparazione del modellato in cera. A sinistra sono indicate alcune dimensioni solitamente consigliate per la spinatura.

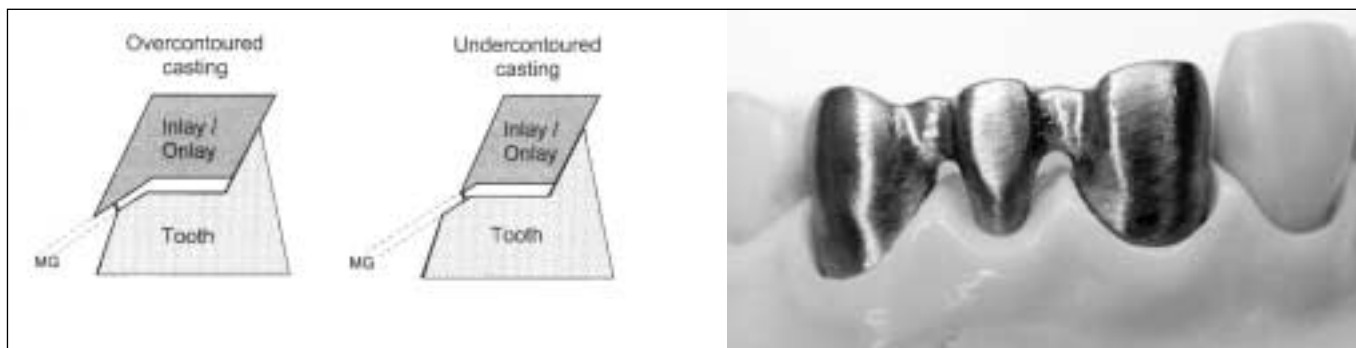


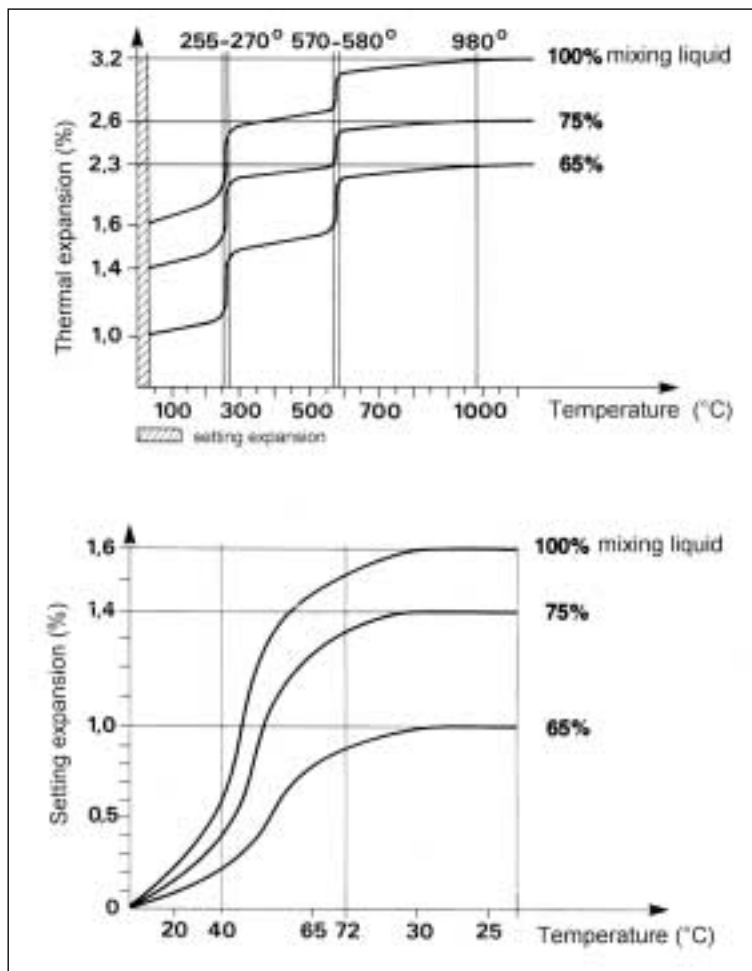
Fig. 2 – Marginal gap. This parameter is a possible accuracy index of the investment casting. On the right: the casting fitted to the pattern.

Fig. 2 – Marginal Gap. Questo parametro è un possibile indice dell'accuratezza della microfusione. A destra la microfusione calzata sul modello.

Fig. 3, Fig. 4 – Graphs summarize the typical trend of the setting expansion of the phosphate bonded casting investment.

During setting, the investment reaches a temperature of about 70°C, then it cools down again. At the end of the process it shows the maximum setting expansion. Later on, during heating into the furnace for wax elimination, it undergoes a thermal expansion that is maintained up until the casting of the alloy. In fact, only a few seconds elapse between the extraction of the dental casting mold from the furnace and the casting of the alloy. Both expansions depend on the percentage of the mixing liquid added to distilled water, used to mix the investment powders. The maximum expansion is obtained by using only the mixing liquid, without water.

Fig. 3, Fig. 4 – I grafici riassumono l'andamento tipico dell'espansione di presa del rivestimento fosfatico. Durante la presa il rivestimento raggiunge una temperatura massima di circa 70 °C, per poi raffreddarsi nuovamente. Al termine del processo presenta la massima espansione di presa. Successivamente, durante il riscaldamento in forno per l'eliminazione della cera, subisce l'espansione termica che viene mantenuta anche al momento della colata della lega al suo interno. Infatti tra l'estrazione del cilindro dal forno e la colata della lega passano pochi secondi. Entrambe le espansioni dipendono dalla percentuale di liquido di miscela che viene aggiunto ad acqua distillata per impastare le polveri di rivestimento. L'espansione massima si ottiene utilizzando il solo liquido di miscela, senza aggiunta di acqua distillata.



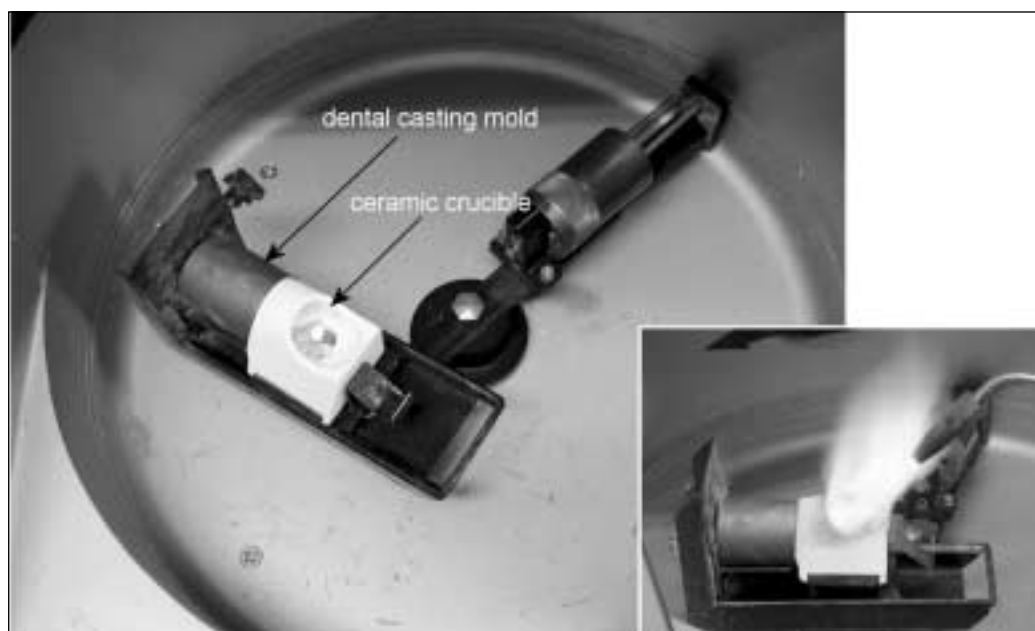


Fig. 5 – The common centrifugal casting machine based on torch heating is still the most widely applied technique in dentistry. The dental casting mold is settled a few seconds before starting the rotation of the knuckled arm and casting the alloy.

Fig. 5 – La classica tecnica di fusione a centrifuga con riscaldamento a cannello è ancora la più diffusa nel settore dentale. Il cilindro viene posizionato pochi secondi prima di avviare la rotazione del braccio snodato e colare la lega.

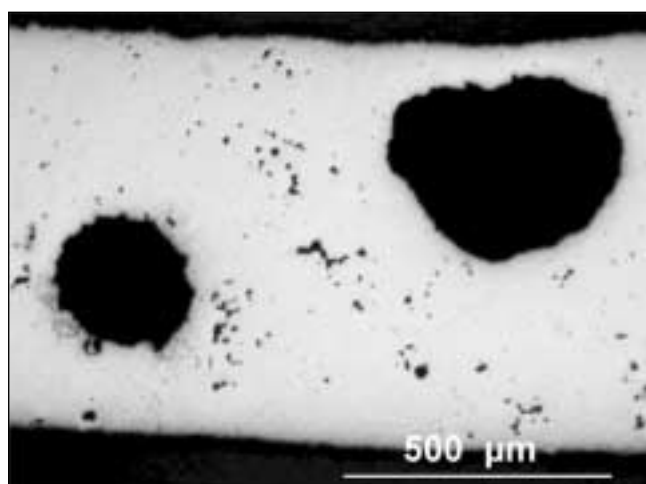


Fig. 6 – Optical metallographic microscope. Alloy 12. Coarse gas porosity. This defect may seriously affect the mechanical performance of the medical device.

Fig. 6 – Microscopio ottico metallografico. Lega tipo 12. Porosità da gas di tipo grossolano. Questo tipo di porosità può essere molto pericoloso per l'integrità meccanica del dispositivo medico.

seconds before the casting starts. Such a technique is still considered as a valid technique for the majority of cases, even though a skilled operator is needed.

Above all, skill is the capability to evaluate the right time to cast. In fact, it is important to estimate the achievement of the optimal casting temperature by eye. This estimation is carried out by looking at the liquid surface of the alloy. Usually, the optimal temperature is achieved just after the vanishing of the oxide film that covers the alloy at the beginning of the liquid state. On the contrary, this does not hold for base metal alloys, since they are always coated by an oxide film even before casting.

Nowadays, furnaces based on new technologies are more and more demanded, for example the induction furnace. In this case it is possible to work under controlled atmosphere, decreasing the alloy oxidation as a result. Moreover, gas pressure casting furnaces are on the market.

Another widely applied heating system is the resistance heating, where the crucible is surrounded by an element heated by the Joule effect.

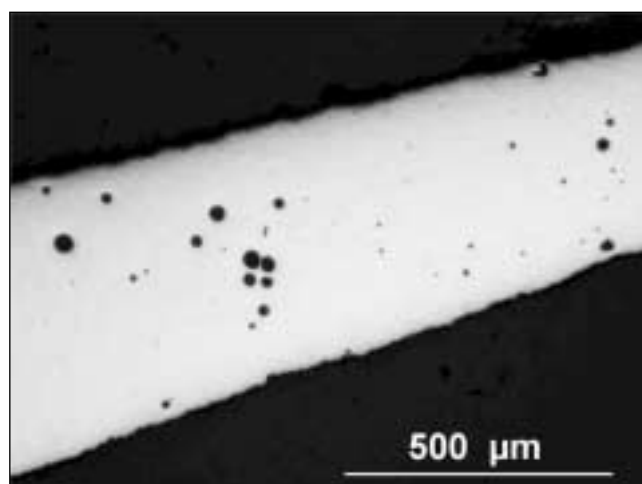


Fig. 7 – Optical metallographic microscope. Alloy 22. Finer gas porosity. This defect does not affect seriously the mechanical properties of the device. However, if it emerges at surfaces contacting the fluids of the oral cavity, it may give rise to corrosion due to differential aeration.

Fig. 7 – Microscopio ottico metallografico. Lega tipo 22. Porosità da gas con dimensioni più contenute. Questo tipo di porosità è meno importante ai fini dell'integrità meccanica del manufatto. Tuttavia, se affiorante alla superficie in contatto con i fluidi della cavità orale, può dare origine a fenomeni di corrosione per aerazione differenziale.

The advantages offered by the modern furnaces are the standardization of parameters -like the casting temperature or the energy transferred to the alloy, and inert gas environments – nitrogen or argon. The main disadvantage is the cost of the apparatus, that is appreciably higher than that of the conventional oxygen-propane torch.

A special aspect of dental alloy casting is the use of scraps coming from previous working. This is crucial for precious metal alloys especially. As in jewelry, the re-processing of scraps is fundamental for economical reasons. However, unlike jewelry alloys, dental alloys may undergo serious degradation upon recasting. This is due to their complex chemical composition, since they contain elements that can oxidize or absorb gases, like Ag and Pd. Additionally, they contain elements that can adhere to the crucible walls due to their extremely high or extremely low melting temperature,

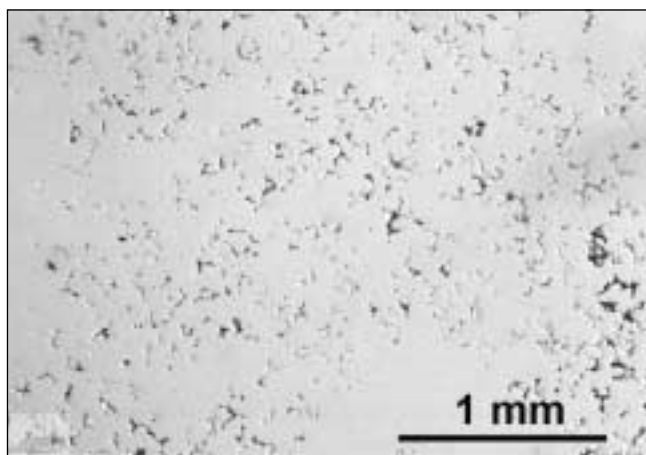


Fig. 8 – Scanning electron microscope. Secondary electrons. Alloy 38. Dispersed shrinkage porosity due to an improper preparation of the sprue former of the wax pattern.

Fig. 8 – Microscopio elettronico a scansione. Elettroni secondari. Lega tipo 38. Diffusa porosità da ritiro causata da un'errata preparazione della spinatura del modellato.

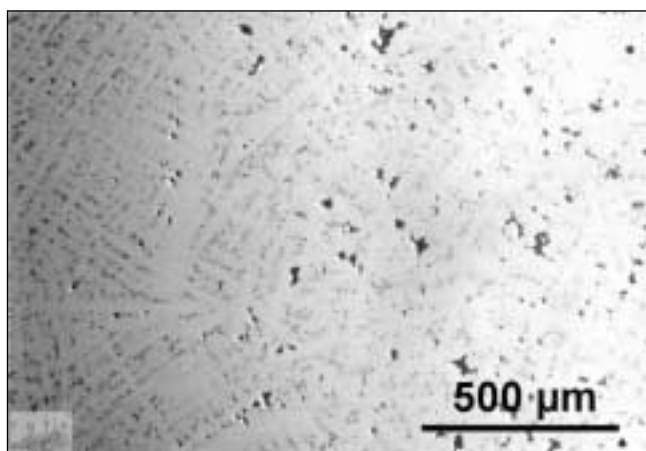


Fig. 9 – Scanning electron microscope. Backscattered electrons. Alloy 46. Shrinkage porosity due to grain refiners depletion, caused by multiple recasting of the alloy, that acquires a dendritic structure.

Fig. 9 – Microscopio elettronico a scansione. Elettroni retrodiffusi. Lega tipo 46. Porosità da ritiro favorita dal depauperamento degli affinatori del grano a causa di rifusioni multiple della lega che assume struttura dendritica.

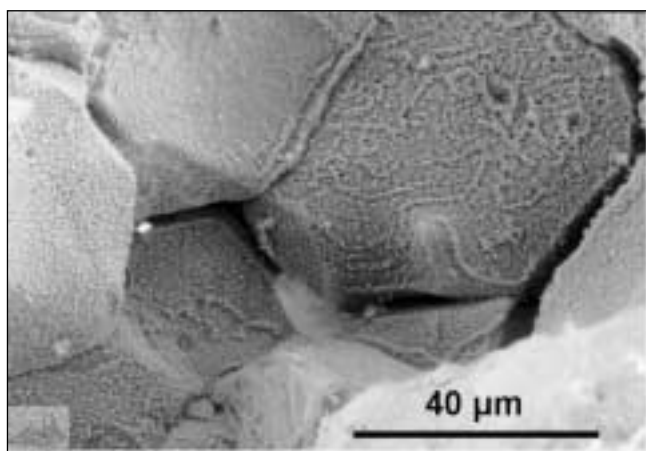


Fig. 10 – Scanning electron microscope. Secondary electrons. Hot tearing in alloy 11. Grain boundary detachment is visible.

Fig. 10 – Microscopio elettronico a scansione. Elettroni secondari. Hot tearing su lega tipo 11. Si notano le scollature al bordo dei grani cristallini.

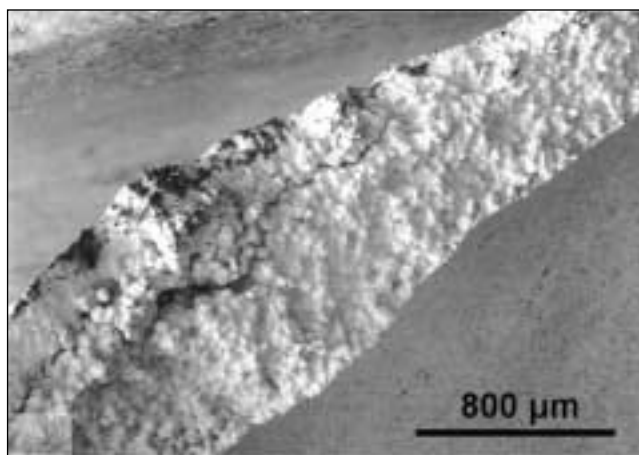


Fig. 11 – Scanning electron microscope. Backscattered electrons. Hot tearing in alloy 24. The intergranular nature of rupture and the brittleness of the alloy are visible.

Fig. 11 – Microscopio elettronico a scansione. Elettroni retrodiffusi. Hot tearing su lega tipo 24. Si nota l'aspetto intergranulare della rottura e la fragilità della lega.

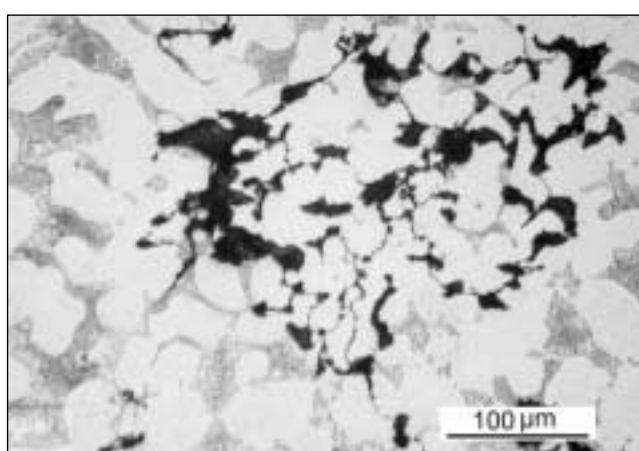


Fig. 12 – Optical metallographic microscope. Oxide membranes in alloy 61.

Fig. 12 – Microscopio ottico metallografico. Membrane di ossidi in una lega del tipo 61.

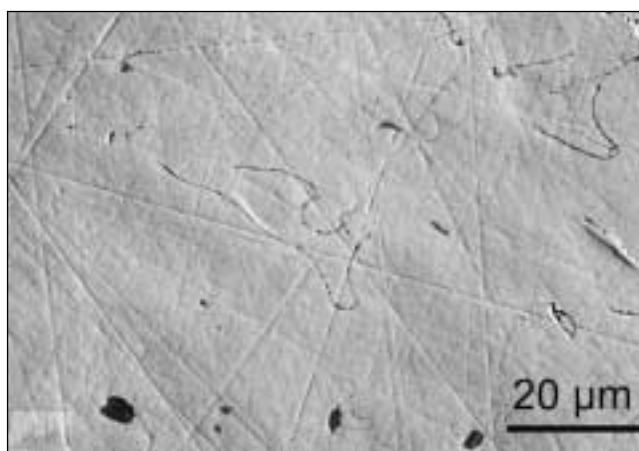


Fig. 13 – Scanning electron microscope. Backscattered electrons. Oxide membranes in alloy 3.

Fig. 13 – Microscopio elettronico a scansione. Elettroni retrodiffusi. Membrane di ossidi in una lega tipo 3.



Fig. 14 – Incomplete device due to a too low casting temperature.

Fig. 14 – Manufatto incompleto a causa della temperatura di colata insufficiente.

like Ir, Ru, Rh and Re in the former case and In, Ga and Sn in the latter. As a consequence, the concentration of these elements in the alloy decreases at each casting.

Zn has a relatively low boiling temperature - about 906°C, and may be partially lost each time the alloy is liquid.

Hence, the profitable reuse of the scraps depends on avoiding alloy damage during the casting process as much as possible. This is achieved by preventing alloy overheating, oxidizing and too much long times in the liquid state in the crucible before casting.

However, it depends on the proper weight balancing of scraps and virgin alloy, above all. The rule of using 50% of virgin alloy at least is always appropriate, even though 60% of virgin alloys is recommended in some cases – especially for alloys with high Pd content, and 40% in alloys with high Au content.

Another important aspect to be considered is the cleanliness of scraps. Mostly, the correct cleaning is performed by milling their surface by tungsten carbide mills followed by sandblasting by aluminium oxide.

FREQUENTLY OBSERVED DEFECTS IN INVESTMENT CASTING OF DENTAL ALLOYS

The variety of defects in investment casting of dental alloys is fairly small. However, defects may show density, dimensions and shape different from case to case. As a consequence, they may have or have no influence at all on the performance of the medical device.

Porosity

The main kinds of porosities in dental alloy investment casting are shrinkage and gas porosity (fig. 6,7,8,9). Shrinkage porosity is mostly due to an improper manufacture of the sprue former of the wax pattern (fig.1) or to an improper positioning of the wax pattern into the dental casting mold.

Particularly, in order to avoid shrinkage porosity as much as possible, it is necessary that the wax pattern is correctly fed by the liquid alloy during casting. Furthermore, the alloy must solidify first in the region corresponding to the wax pattern and then in the sprue former.

The correct sequence of solidification depends on the positioning of the wax pattern into the dental casting mold. Gas porosity may be caused by different factors, like the use of flames with excessive gas pressure, turbulence during ca-



Fig. 15 – Typical fins on a portion of the former sprue, due to the investment breaking during pre-heating of the dental casting mold.

Fig. 15 – Caratteristiche pinne su parte della spinatura, causate dalla rottura del rivestimento durante le fasi di preriscaldamento del cilindro.

sting due to wrong dimensions of the sprue former or erroneous runners orientation. This defect may be present even in castings obtained by pressure casting furnaces.

Hot tearing

Hot tearing (fig. 10,11) occurs during the initial solidification stages of the alloy into the dental casting mold.

It is caused by grain boundary phases with solidification temperatures lower than that of the grain matrix. These phases are promoted by grain boundary macrosegregation.

The material is inclined to break along these interfaces, since the mechanical resistance of these phases is not enough to counterbalance all possible tensile stresses due to the different thermal contraction of the investment and of the alloy.

The problem can be solved by trial, that is by varying the preheating temperature of the dental casting mold and the casting temperature, until the defect disappears. A good practice is to lower the preheating temperature of the dental casting mold and the casting temperature as much as possible. The defect may be present even if the grain is refined properly. However, if the loss of grain refiners leads to an excessive grain size, the defect may appear unexpectedly. This defect is particularly critical because ruptures may be partial and not observable before the insertion of the device into the oral cavity.

Alloy oxidation

The use of scraps coming from previous working or excessive alloy oxidation during investment casting may lead to oxides accumulation. Such oxides are often membranes, that show a filiform aspect when observed on a metallographic section (fig. 12,13). These microstructural defects may cause ruptures during service in the oral cavity, since they act as fatigue rupture initiators.

Defects due to improper preparation of the dental casting mold and to improper casting temperature

This kind of defects is similar to that observed in investment casting for jewelry. If the casting temperature is too low, material lacking in the device may occur (fig. 14). The same effect is produced if the pre-heating temperature of the dental casting mold is too low. If the preparation procedure of the dental casting mold is incorrect, for example if the pre-heating rate of the investment is too high, ruptures of the investment may occur. In this case the alloy shows the typical fins along the edges (fig. 15).

CONCLUSIONS

The main aspects that the investment casting for dentistry must face relate to the wide variety of available alloys and to the environment the device should be applied to, that is the oral cavity.

The accuracy of reproduction of the wax pattern is the main task of the technician. Furthermore, it is necessary that the alloy microstructure is not altered by the casting process and that microstructural defects are as few as possible, in order to achieve the highest device reliability in relation to its employment in the oral cavity.

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**LA MICROFUSIONE AL SERVIZIO DEL CAVO ORALE:
ASPETTI PECULIARI E PROBLEMATICHE**

PAROLE CHIAVE: metalli preziosi, biomateriali, difetti, frattura, solidificazione, fonderia, getti, difettologia, metallografia, microscopia elettronica, proprietà, selezione materiali

La microfusione a cera persa in ambito odontoiatrico presenta due aspetti fondamentali che ne caratterizzano la peculiarità.

Il primo aspetto è nell'obiettivo stesso del processo, finalizzato alla realizzazione di un dispositivo medico su misura. Infatti ci si prefigge di ottenere un manufatto che si adatti nel miglior modo possibile alle caratteristiche geometriche proprie della cavità orale del paziente. Ne consegue che la microfusione realizza sempre pezzi unici per forma e dimensioni, essendo così esclusa una produzione in serie come può viceversa avvenire nel settore orafa.

Il secondo aspetto deriva direttamente dal primo e riguarda la precisione che il dispositivo deve avere al fine di essere impiegato in cavità orale. La precisione da ottenere è fondamentale non solo per la funzionalità meccanica del manufatto, ma anche per il rispetto dell'ambiente orale e per il mantenimento dei requisiti essenziali del dispositivo nel tempo. Ad esempio, è necessario che l'introduzione in cavità orale non alteri gli equilibri masticatori, non determini alterazioni dei tessuti, non introduca spazi e ambienti dove la flora batterica o fenomeni di degrado dei materiali (corrosione, abrasione) possano portare rapidamente alla necessità di rimuovere lo stesso dispositivo. Una caratteristica importante per la durata del dispositivo è l'ampiezza della intercapedine di chiusura marginale tra il moncone di un dente e la microfusione inserita sopra di esso. In questo caso si raccomanda che questa intercapedine non superi l'ampiezza di poche decine di micron ($30 \div 40$) anche se alcuni ritengono tollerabili ampiezze fino a $100 \mu\text{m}$. Tanto più questa chiusura è precisa, tanto più il paziente è in grado di mantenerla pulita, rimuovendo la placca batterica.

L'obiettivo da raggiungere è quindi l'assenza di differenze significative tra il modello in cera e la fusione ottenuta. E' comunque difficile esprimere questo concetto in termini quantitativi, sia per l'unicità della geometria di ogni fusione che per le valutazioni soggettive di questa precisione da parte del medico preposto all'inserimento del dispositivo in cavità orale.

Occorre ricordare che i dispositivi di cui si parla sono realizzati abbinando ad un substrato metallico, ottenuto per microfusione, un rivestimento estetico in resina o in ceramica. L'accuratezza con cui il substrato metallico duplica il modello in cera si ottiene con il bilanciamento tra le deformazioni che cera, rivestimento e lega subiscono durante l'intero processo. In particolare, si ricorda la contrazione della cera durante la preparazione del modellato (circa lo 0,2%), la sua espansione durante la messa in rivestimento prima di bruciare, l'espansione del rivestimento durante l'indurimento ($0,3 \div 1,5$ %) e la contrazione della lega in solidificazione ($1,6 \div 2$ %).

L'ottimizzazione del processo è quindi frutto di un lavoro empirico che può portare a soluzioni diverse, anche a causa

della grande differenza e tipologia di prodotti impiegati per ognuna delle quattro seguenti categorie di materiali: cere, rivestimenti gessosi o a legante fosfatico, leghe. Per quanto riguarda le tecniche di messa in rivestimento, si ricorda in particolare la possibilità di preparare il cilindro con contenitore metallico oppure senza di esso, per l'esecuzione della tecnica ad espansione libera; nonché la possibilità di regolare l'espansione del rivestimento a legante fosfatico mediante un'opportuna miscelazione dell'acqua di impasto con una soluzione di silice colloidale.

Si può comunque affermare che il livello di accuratezza raggiunto nella precisione delle microfusioni odierne è stato ottenuto mediante il miglioramento delle caratteristiche dei materiali e delle tecniche di fusione, tra le quali rimangono di primaria importanza la tecnica di colaggio centrifugo con fusione della lega a cannello, a induzione o a resistenza elettrica, e la tecnica di pressofusione.

Il primo utilizzo del processo a cera persa per la microfusione in ambito odontoiatrico risale al 1891 ad opera di Martin, che effettuò colate di oro in uno stampo preparato a cera persa. Successivamente vennero sviluppati nuovi materiali fino ad arrivare alle moderne leghe di metalli preziosi per rivestimento estetico in resina. Nel frattempo, a partire dal 1950, ad opera di Weinstein, vennero realizzate leghe in metalli preziosi per rivestimento estetico in ceramica. Oggi sono in commercio diverse centinaia di leghe appartenenti a queste due categorie, alle quali si aggiungono le leghe a base nichel-cromo, cromo cobalto ed il titanio e sue leghe.

E' possibile suddividere in famiglie le varie leghe dentali, sulla base del loro impiego per rivestimento estetico in resina o in ceramica e, all'interno di queste due ripartire le diverse leghe sulla base della loro composizione chimica. In generale, tra le leghe di metalli preziosi si individuano quelle ad alto titolo aureo (80%), a medio titolo aureo (50% con o senza palladio), a basso titolo aureo (20%), ad alto titolo di palladio (70% circa) con o senza oro, al palladio-argento con o senza oro.

Il grande numero di leghe utilizzate è un'altra peculiarità del processo di microfusione applicato al settore odontoiatrico. Spesso le leghe non sono monofasiche ma presentano fasi e costituenti strutturali diversi. Da ciò conseguono comportamenti a solidificazione differenti, anche dovuti al variare della densità, dell'intervallo di fusione, della diversa propensione all'ossidazione durante la fusione e la colata.

Nonostante l'enorme possibilità di combinazioni tra leghe, apparecchiature per microfusione e rivestimenti, la tipologia dei difetti di fusione è relativamente modesta. Tra questi, i più dannosi sono quelli interni al getto, che non possono essere individuati all'osservazione visiva. In particolare, le porosità possono essere causa di rotture differite in cavità orale, anche in associazione a fenomeni di fatica. La perdita dei raffinatori del grano determina un degrado nelle caratteristiche meccaniche e di resistenza alla corrosione della lega.

In conclusione, la microfusione in campo odontoiatrico, nonostante lo sviluppo continuo delle tecniche e dei materiali, risente ancora di un buon grado di empirismo, favorito dall'unicità di forme e geometrie proprie di ogni lavorazione e dalla grande tipologia di leghe a disposizione del tecnico.